Image quality improvement in megavoltage cone beam CT using an imaging beam line and a sintered pixelated array system

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(Received 25 March 2011; revised 31 August 2011; accepted for publication 12 September 2011; published 17 October 2011)

Purpose: To quantify the improvement in megavoltage cone beam computed tomography (MVCBCT) image quality enabled by the combination of a 4.2 MV imaging beam line (IBL) with a carbon electron target and a detector system equipped with a novel sintered pixelated array (SPA) of translucent Gd₂O₂S ceramic scintillator. Clinical MVCBCT images are traditionally acquired with the same 6 MV treatment beam line (TBL) that is used for cancer treatment, a standard amorphous Si (a-Si) flat panel imager, and the Kodak Lanex Fast-B (LFB) scintillator. The IBL produces a greater fluence of keV-range photons than the TBL, to which the detector response is more optimal, and the SPA is a more efficient scintillator than the LFB.

Methods: A prototype IBL + SPA system was installed on a Siemens Oncor linear accelerator equipped with the MVision™ image guided radiation therapy (IGRT) system. A SPA strip consisting of four neighboring tiles and measuring 40 cm by 10.96 cm in the crossplane and inplane directions, respectively, was installed in the flat panel imager. Head- and pelvis-sized phantom images were acquired at doses ranging from 3 to 60 cGy with three MVCBCT configurations: TBL + LFB, IBL + LFB, and IBL + SPA. Phantom image quality at each dose was quantified using the contrast-to-noise ratio (CNR) and modulation transfer function (MTF) metrics. Head and neck, thoracic, and pelvic (prostate) cancer patients were imaged with the three imaging system configurations at multiple doses ranging from 3 to 15 cGy. The systems were assessed qualitatively from the patient image data.

Results: For head and neck and pelvis-sized phantom images, imaging doses of 3 cGy or greater, and relative electron densities of 1.09 and 1.48, the CNR average improvement factors for imaging system change of TBL + LFB to IBL + LFB, IBL + LFB to IBL + SPA, and TBL + LFB to IBL + SPA were 1.63 ($p < 10^{-7}$), 1.64 ($p < 10^{-13}$), and 2.66 ($p < 10^{-9}$), respectively. For all imaging doses, soft tissue contrast was more easily differentiated on IBL + SPA head and neck and pelvic images than TBL + LFB and IBL + LFB. IBL + SPA thoracic images were comparable to IBL + LFB images, but less noisy than TBL + LFB images at all imaging doses considered. The mean MTFs over all imaging doses were comparable, at within 3%, for all imaging system configurations for both the head- and pelvis-sized phantoms.

Conclusions: Since CNR scales with the square root of imaging dose, changing from TBL + LFB to IBL + SPA reduces the imaging dose required to obtain a given CNR by factors of 0.38 and 0.37, respectively. MTFs were comparable between imaging system configurations. IBL + SPA patient image quality was always better than that of the TBL + LFB system and as good as or better than that of the IBL + LFB system, for a given dose. © 2011 American Association of Physicists in Medicine. [DOI: 10.1118/1.3651470]

Key words: megavoltage cone beam CT, image quality

I. INTRODUCTION

Megavoltage cone beam computed tomography (MVCBCT) as implemented in the Siemens (Erlangen, Germany) MVision™ system, generates 3D patient images for image guided radiation therapy (IGRT) using conventional equipment: an electronic portal imaging device (EPID) with amorphous silicon (a-Si) detectors and the same 6 MV, flattened, treatment beam line (TBL) used to deliver the therapeutic radiation dose. The simplicity of the system provides the benefit of straightforward quality assurance since the imaging and treatment isocenters are shared, reducing equipment and maintenance costs. The increased beam energy of MVCBCT relative to kilovoltage (kV) cone beam (kVCBCT) and kV helical CT (kVCT) imaging systems also
provides the benefit of reduced metal artifacts in the images.\textsuperscript{2,3} Since the TBL is modeled in the treatment planning system, daily MVCBCT imaging dose, which can approach 5% of the total prescription dose for prostate patients,\textsuperscript{4–6} can be managed through calculation and incorporation into treatment plans.\textsuperscript{4–6} This strategy satisfies the suggestion of AAPM Task Group 75\textsuperscript{8} to manage imaging dose in IGRT. Image quality at a given imaging dose is typically better for kVCBCT imaging systems than TBL-based MVCBCT\textsuperscript{9} since MeV-range x-ray image contrast is dominated by Compton, rather than photoelectric, interactions.\textsuperscript{10} and the EPID response is typically optimal for keV-range x-rays.\textsuperscript{11} Improving MVCBCT image quality is the subject of the current work, which focuses on the Siemens MVision\textsuperscript{TM} MVCBCT system.

As MVCBCT image quality improves, the clinical adoption of MVCBCT-based dose-guided radiation therapy\textsuperscript{12,13} becomes increasingly attractive. MVCBCT image quality for the MVision\textsuperscript{TM} system has been improved through the introduction of an unflattened 4.2 MV imaging beam line (IBL) with a low atomic number (Z) target material such as graphite\textsuperscript{11,14,15} or diamond,\textsuperscript{16} rather than the tungsten electron target used for the TBL. Low-Z MVCBCT imaging has also been implemented on a Varian (Palo Alto, CA) accelerator using aluminum\textsuperscript{17,18} and beryllium\textsuperscript{18} electron targets and on an Elekta (Stockholm, Sweden) linear accelerator with a carbon electron target.\textsuperscript{19} The use of low-Z electron targets for MVCBCT improves image quality, as may be expected from previous studies that demonstrated that MeV-range x-ray beams generated from low-Z electron targets improve the quality of portal images.\textsuperscript{20–22} IBL-based MVCBCT also retains the benefit that imaging dose can be incorporated into the treatment plans using a standard treatment planning system.\textsuperscript{6}

MVCBCT image quality can also be improved by replacing the standard Gd$_2$O$_2$S Lanex Fast-B (LFB) (Carestream, Rochester, NY) scintillator in the EPID with a thicker scintillator that is segmented by septa of reflective epoxy into pixels of the same pitch as the underlying a-Si detectors.\textsuperscript{23} The increased scintillator thickness increases the detector quantum efficiency, but potentially reduces resolution due to a wider light-spread function. To address this issue, the reflective septa reduce the spread of optical photons to neighboring pixels, reducing a loss in spatial resolution.\textsuperscript{24,25} Segmented scintillators composed of crystalline materials such as CsI(Tl) (Refs. 23–25) and BGO,\textsuperscript{24,25} as well non-crystalline materials such as Gd$_2$O$_2$S:Tb powder,\textsuperscript{26} have been developed to optimize the tradeoff between detector quantum efficiency and loss in spatial resolution. A disadvantage of the crystalline scintillators is that they are costly and complex to manufacture without defects.\textsuperscript{24,26} The powder-based segmented Gd$_2$O$_2$S:Tb systems tend not to outperform the conventional Kodak LFB scintillator due to the presence of Swank noise, which arises due to depth-dependent gain in thick, opaque scintillators, and decreases obtainable detective quantum efficiency (DQE).\textsuperscript{26,27}

In this work, a segmented scintillator composed of Gd$_2$O$_2$S ceramic,\textsuperscript{28} called a sintered pixelated array (SPA), is combined with IBL-based MVCBCT to improve image quality. The SPA is designed to (i) be less expensive and complex to manufacture than traditional segmented crystalline scintillators and (ii) reduce the deleterious effects of Swank noise on image quality relative to that obtained with opaque, powder-based Gd$_2$O$_2$S.\textsuperscript{26} Issue (i) is addressed with an automated manufacturing process that entails micromachining pixel-separating grooves into a solid slab of Gd$_2$O$_2$S ceramic, then injecting the grooves with a reflective epoxy under high pressure. The Gd$_2$O$_2$S ceramic slab is formed by sintering a powder mixture of Gd$_2$O$_2$S that is doped for fast light decay using a proprietary formula. Since the SPA is less than 1 cm thick, at 1.8 mm, the pixels do not need to be focused toward the beam source,\textsuperscript{24} thus the manufacture of pixels of only one shape is sufficient. The SPA is homogeneous and addresses issue (ii) since it is yellow-green translucent, which reduces the effects of Swank noise relative to a powder-based material such as the LFB scintillator. The SPA also has a 59% higher density than the LFB material, which increases the efficiency of the scintillator per unit thickness.

In this work, a series of phantom studies for quantitative image quality assessment are presented that compare the baseline imaging system, TBL + LFB, to two imaging system configurations: IBL + LFB and IBL + SPA. The first head and neck (H&N), thoracic, and pelvic patient images acquired at multiple different imaging doses with the three system configurations are also presented.

II. MATERIALS AND METHODS

II.A Beam, detector, and reconstruction configurations

All MVCBCT images were acquired on Siemens Oncor linear accelerators equipped with the MVision\textsuperscript{TM} IGRT system\textsuperscript{2} using either a TBL or an IBL. The IBL installation process was described in detail by Faddegon et al.\textsuperscript{11} The LFB and SPA systems are standard and modified, respectively, versions of the Perkin Elmer Optoelectronics (Weisbaden, Germany or Santa Clara, CA) AN9 flat panel imager. Each flat panel imager contained a 1024$\times$1024 matrix of a-Si detectors of 0.4 mm$\times$0.4 mm area each, thus the detector area was 40.96 cm$\times$40.96 cm. Images were acquired at source-to-imager-distance of 145 cm, source-axis-distance of 100 cm, and the maximum imaging beam field size is 27.4 cm$\times$27.4 cm.

A cross sectional diagram of the LFB system is shown in Fig. 1(a). In the downstream-to-upstream imaging beam direction, the LFB system was composed of glass substrate, an a-Si photodiode layer, a 0.26 mm thick LFB scintillator with an areal density of 0.134 g/cm$^2$, a 1 mm thick copper buildup plate, 3 mm of empty space, and a 0.75 mm thick aluminum cover. The IBL + LFB system under consideration is the same configuration used for the Siemens kView\textsuperscript{TM} system.

A cross sectional diagram of the SPA system is shown in Fig. 1(b). In the downstream-to-upstream imaging beam direction, the SPA system was composed of the same glass
substrate and photodiode system as the conventional imager, a 1.8 mm thick segmented SPA scintillator with a fill factor of 0.64 and an areal density of 1.32 g/cm², a 2.46 mm thick aluminum slab that held the SPA in place, and the same 0.75 mm thick aluminum cover as the LFB system. The SPA system’s scintillator consisted of four separate SPA tiles, shown in Fig. 2, which lay across the photodiode detector surface, forming a strip of 40 cm length in the direction of gantry rotation (crossplane) and 10.96 cm width in the longitudinal (inplane) direction. The SPA tiles were fixed to the aluminum plate with double-sided adhesive tape, immobilizing them with respect to the plate. To prevent motion of the plate with respect to the photodiode array, the plate was designed to fit snugly within the internal housing of the flat panel imager. The screws used to attach the radiation shields at the edges of the flat panel chassis were loosened in order to insert the plate. When the screws were tightened, the plate was effectively clamped in place.

The SPA pixels were not optically coupled to the a-Si detectors, thus the light mediation between the different refractive indices of the two materials was nonideal. A rigorous test was not performed to test the alignment between the SPA and detector array, therefore perfect alignment was not expected. Central buckling of the aluminum plate could have introduced a gap between the SPA and the detector array, which could cause some variation in SPA position relative to the photodiode array with gantry angle. A process for optimizing the optical coupling with the SPA and ensuring excellent immobilization for all gantry angles is under development.

II.B Upper bound on DQE values

In order to quantify the expected differences in image quality with each imaging system configuration, an upper bound on detective quantum efficiency (DQE) for each system was calculated by Monte Carlo modeling. The imaging systems with cross sections shown in Fig. 1 were modeled as 12.8 × 12.8 mm (32 × 32 pixel) area slabs using the user code DOSXYZnrc29 of the Monte Carlo simulation package EGSnrc.30 The TBL and IBL beams were modeled as 3.2 × 3.2 mm square cross sections of normally-incident photons using the same TBL and IBL beam parameters employed by Maltz et al.31 The multi-pixel model was large enough to account for the effects of photons incident at adjacent pixels on the response of the pixel of interest as well as scatter originating within the detector body. The photon and electron energy transport cutoffs for the Monte Carlo simulations were 10 and 512 keV, respectively.

An upper bound on DQE was calculated based on the principle that perfect detection, or a DQE of 100%, is obtained when all of the energy in the incident x-ray beam is deposited in the detector elements. Since an upper bound rather than an exact DQE was desired, the following simplifying assumptions were made for the calculation: (i) all kinetic energy deposited inside the scintillator was converted

![Area zoomed](image-url)
to optical photons, and (ii) all optical photons generated in the scintillator reached and were absorbed in the detector array. The upper bound on DQE for each imaging system configuration, $DQE_{UB}$, was calculated as:

$$DQE_{UB} = F \int_{0}^{\infty} \frac{\phi(E)e^{-\mu(E)}\mathcal{F}(E)dE}{\int_{0}^{\infty} \phi(E)e^{-\mu(E)}dE}, \quad (1)$$

where $F$ is the scintillator fill factor, $E$ is x-ray energy, $\phi(E)\exp[-\mu(E)t]$ is the x-ray beam spectrum corrected for transmission through the phantom of water-equivalent thickness $t$ and with attenuation coefficient $\mu(E)$, $f(E)$ is the Monte Carlo calculated fraction of energy of a monoenergetic photon beam of energy $E$ that is deposited in the detector array.

II.C Phantoms

Two phantoms were used to quantify image quality as a function of imaging dose and imaging system. The first [Fig. 3(a)] was a head-sized cylindrical phantom referred to as EMMA and described in detail by Gayou and Miften. Briefly, the EMMA phantom is a 20 cm diameter cylindrical phantom with a 2 cm thick high contrast region containing 2 cm diameter cylindrical contrast inserts with relative (to water) electron densities (rEDs) of 0.001 (air), 1.09 (inner bone), 1.17 (acrylic), and 1.48 (bone). The EMMA phantom also contains a 2 cm thick acrylic spatial resolution evaluation region with six groups of air bar patterns with spatial frequencies of 0.67, 1.0, 1.5, 2.0, 2.5, and 3.0 line pairs per cm. The contrast and spatial resolution regions were used to determine the contrast-to-noise ratio (CNR) and the modulation transfer function (MTF), respectively, of each imaging configuration as described in Sec. II F.

The second phantom considered was the pelvis-sized large EMMA phantom (LEMA), shown in Fig. 3(b). LEMA is the EMMA phantom surrounded by an acrylic annulus with an outer diameter of 35 cm. Since the same regions of the EMMA phantom were present in both phantom images, the same image quality analysis techniques were applied to both phantoms.

II.D Patients

MVCBCT images of H&N, thoracic, and pelvic (prostate) cancer patients were acquired as part of an Institutional Review Board-approved pilot study. Each patient was imaged with the TBL + LFB, IBL + LFB, and TBL + SPA system configurations at multiple doses in order to determine the expected image quality for each imaging beam, imaging system, and imaging dose. Imaging dose was incorporated into the treatment plans for all cases, and the imaging sessions were designed in a manner that prevented the patients from receiving a cumulative imaging dose in excess of that reported by the treatment planning system at the time the plan was approved.

II.E Imaging parameters

Phantom and patient images were reconstructed at slice thicknesses of 1 and 3 mm, respectively, and had 1 mm × 1 mm resolution in the axial (transverse) plane. Phantom images were analyzed and displayed at 3 mm slice thickness using slice averaging. The phantom images were reconstructed with smaller slice thicknesses to ensure that enough image samples were present in each analysis region to calculate the uncertainties associated with the image quality metrics, as discussed in Sec. II G. H&N and EMMA images were reconstructed with the “edge enhancing head and neck” protocol. Thoracic images were reconstructed with the “edge enhancing” protocol. Pelvic and LEMA images were reconstructed with the “smoothing pelvis” protocol. All reconstruction protocols were the same MVision protocols used at the University of Iowa Hospitals and Clinics (UIHC).

Phantom images were acquired at doses of 3, 5, 10, 15, 30, 45, and 60 cGy, where the reported imaging dose is that which would be delivered to the point of maximum dose in a water phantom on the central beam axis, from a fixed IBL or TBL beam, with a field size of 10 cm × 10 cm defined at a distance of 100 cm from the source. All patients were imaged at 3, 5, and 10 cGy, and the pelvic (prostate) cancer patient was also imaged at 15 cGy. The standard H&N, thoracic, and pelvic imaging doses at the UIHC are 5, 10, and 15 cGy, respectively.

For the SPA system, junctions between SPA tiles resulted in missing data that caused ring artifacts in the reconstructed images. The Laplacian solution ring artifact correction algorithm of Nelms et al. was applied to the projection images prior to reconstruction to reduce the impact of the artifacts on image quality.

II.F Image display

All images shown in this work are displayed at 3 mm slice thickness in a manner designed to prevent unfairly biasing the visible image noise in favor of any one imaging system. This was accomplished for the patient images by defining the displayed intensity window as twice the difference between the average intensities in 1 cm² bone and soft tissue regions of interest (ROIs) in the image. The displayed intensity level was the average of the average bone and soft
tissue intensities in the corresponding ROIs. The EMMA and LEMMA phantom images were displayed at the same window as the H&N and pelvic patient images, respectively, and the level for each phantom image was recalculated using the same method as for the patient images.

II.G Image quality quantification

II.G.1 CNR

The CNR was calculated for all phantom images in the contrast regions with rEDs of 1.09 and 1.48 with the following method previously for MCBCT analysis by Morin et al.:\(^{34}\)

\[
\text{CNR} = \frac{\text{CR}_{\text{mean}} - \text{BG}_{\text{mean}}}{(\text{CR}_{\text{std}} + \text{BG}_{\text{std}})/2}.
\]

In Eq. (2), CR\(_{\text{mean}}\) and CR\(_{\text{std}}\) are the mean and standard deviation, respectively, of the image intensities in the contrast regions, and BG\(_{\text{mean}}\) and BG\(_{\text{std}}\) are the mean and standard deviation, respectively, of the image intensities in background regions neighboring each contrast region. The ROIs used in the CNR analysis are shown in Fig. 4(a), and were defined in a manner that avoided ring artifacts induced by the SPA tile junctions in the IBL + SPA images. Although the TBL + LFB and IBL + LFB images did not have ring artifacts, consistency was ensured by using the same ROIs, pixel-for-pixel, for all phantom images.

Theoretical CNR ratios between the two imaging system configurations, \(b\) and \(a\), were calculated using the DQE\(_{\text{UB}}\) values calculated with Eq. (1) as follows:

\[
\frac{\text{CNR}_b}{\text{CNR}_a} = \frac{\sqrt{\text{DQE}_{\text{UB},b}(0)}}{\sqrt{\text{DQE}_{\text{UB},a}(0)}}
\]

where the additional subscripts associated with CNR and DQE\(_{\text{UB}}\) identify the imaging system configuration under consideration. Equation (3) follows from Eq. (66) of Barrett et al.,\(^{35}\) which states that the square of signal-to-noise ratio, and thus the square of CNR, is proportional to quantum efficiency.

II.G.2 MTF

The spatial resolution of the imaging system configurations was quantified with the MTF. The MTF was calculated for each phantom image using the method of Droege and Rzeszotarski\(^{36}\) with some minor modifications\(^{37}\) described by Nelms et al.\(^{33}\) The MTF was calculated as:

\[
\text{MTF}_n = \pi \frac{2}{M_0} \sqrt{V_{\text{BP}}(f_n) - \frac{1}{9} V_{\text{BP}}(3f_n)},
\]

where \(V_{\text{BP}}(f_n)\) represents the difference between the image and noise variance of the bar pattern at frequency, \(f_n\). The ROIs used for the MTF calculations are shown in Fig. 4(b). Image noise within an ROI on a given slice was calculated as the standard deviation of the image intensity difference between neighboring slices, divided by \(\sqrt{2}\), as used by Rajapakshe.\(^{17}\) \(M_0\) is the bar pattern amplitude, given by half the difference between the mean intensity value of the phantom in a region with no bar patterns and the mean intensity value of the air region on an image. \(V_{\text{BP}}(3f_n)\) was determined by linear interpolation at frequencies for which bar patterns were not available.\(^{38}\) EMMA and LEMMA both contain eleven frequency bar patterns but the five bar patterns above the cutoff frequency of 0.3 mm\(^{-1}\) were omitted from MTF calculations for two reasons. First, sensitivity to slight image intensity variation becomes significant due to the small number of voxels in the bar patterns above 0.3 mm\(^{-1}\). Second, experiments have shown that 0.3 mm\(^{-1}\) is the spatial frequency limitation of the TBL + LFB imaging configuration.\(^{39}\)

The reported MTF and CNR values are the average of five (of eleven) central 3 mm slices in the spatial resolution region and the contrast region, respectively. For the MTF calculation, the five 3 mm slices with the lowest noise (in the background region) plus variance (in the bar pattern region) contributed to the reported MTF values. For the CNR calculation, the five slices that resulted in the lowest CNR standard deviation contributed to the mean and standard deviation CNR values. This technique was used to lower the sensitivity of the CNR and MTF calculations to outliers due to ring artifacts produced by the SPA tile junctions, as described in Sec. II.E.

II.G.3 Statistical analysis

The statistical significance of CNR and MTF changes between a given pair of imaging systems was determined by calculating \(p\)-values. The \(p\)-values are the minimum probabilities of mistakenly concluding that the average percentage change of CNR (over all doses) or MTF (over all frequencies) between two imaging systems was nonzero when there was actually no change. The \(p\)-values were calculated using one-sided t-tests.\(^{40}\) An MTF or CNR change was considered statistically significant if the corresponding \(p\)-value was less than 0.05.

III. RESULTS

The calculated DQE\(_{\text{UB}}\) values for each imaging system as a function of the equivalent water thickness of the imaged object are listed in Table I.
Figure 5 shows the same axial slice containing the contrast inserts for the EMMA and LEMMA phantom images for three of the seven imaging doses considered. H&N (axial), pelvic (axial), and thoracic (coronal) patient images are shown in Figs. 6–8, respectively.

The CNR and MTF plots for the phantom images are shown in Figs. 9 and 10, respectively. Analyses of the CNR and MTF changes with imaging system configuration and phantom sizes are shown in Tables 2 and 3, respectively. The lines and error bars for the MTF plots in Fig. 10 correspond to the average and standard deviations of the MTF values, respectively, calculated over all imaging doses considered.

For each phantom, Table IV lists the theoretical, measured, and percentage differences in CNR ratios for imaging system configuration change. The theoretical CNR change data in Table IV were obtained from Table I using Eq. (3).

Table I. Detective quantum efficiency upper bound (DQE UB) calculations. The 20 and 35 cm water thickness cases are representative of the head-sized (EMMA) and pelvis-sized (LEMMA) phantoms, respectively.

<table>
<thead>
<tr>
<th>Water thickness (cm)</th>
<th>DQE UB</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>TBL + LFB (%)</td>
</tr>
<tr>
<td>0</td>
<td>1.07</td>
</tr>
<tr>
<td>20</td>
<td>0.69</td>
</tr>
<tr>
<td>35</td>
<td>0.57</td>
</tr>
</tbody>
</table>

Notes: TBL + LFB: Treatment beam line (TBL) equipped with Lanex Fast-B scintillator (LFB), IBL + LFB: Imaging beam line (IBL) with LFB, IBL + SPA: IBL with sintered pixelated array (SPA) scintillator.

IV. DISCUSSION

IV.A DQE upper bound calculations

The DQE upper bound calculations listed in Table I indicate ~1% and 22% DQE upper bounds for TBL + LFB and IBL + SPA, respectively, which are consistent with published DQE(0) data for 6 MV beams combined with conventional phosphor screen-based EPIDs41 and a thick (40 mm) segmented CsI(Tl) crystalline scintillator.42 Although the imaging system configurations considered in the current study do not exactly match those for which published DQE(0) in the literature, the calculated values in Table I appear reasonable.

III.B Phantom images

It is evident in all IBL + SPA images that the ring artifact correction algorithm did not entirely remove the ring artifacts due to the SPA tile junctions. Inspection of the phantom images in Fig. 5 reveals a noise reduction for all imaging system changes and all doses.

IV.C Patient images

The H&N images in Fig. 6 show clear improvement from TBL + LFB to IBL + LFB and slight improvement from IBL + LFB to IBL + SPA. The pelvic images in Fig. 7 show a clear noise reduction from TBL + LFB to IBL + LFB to IBL + SPA. IBL + SPA images provide improved soft tissue differentiation in the pelvis relative to the TBL + LFB and IBL + LFB images. The thoracic images in Fig. 8 show reduced noise from TBL + LFB to IBL + LFB, but no obvious improvement is seen from the IBL + LFB images to the IBL + SPA images. Little qualitative improvement in IBA + SPA thoracic image quality was visible beyond an imaging dose of 5 cGy.

IV.D CNR

Figure 9 shows that each imaging system change results in an obvious CNR change for imaging doses of 3 cGy or greater. Table II indicates that each imaging system change provides a statistically significant CNR improvement factor ($p < 0.05$), for each contrast insert and each phantom. When all phantoms and all contrast inserts are considered together, the CNR average improvement factors for imaging system changes of TBL + LFB to IBL + LFB, IBL + LFB to IBL + SPA, and TBL + LFB to IBL + SPA were 1.63 ($p < 10^{-7}$), 1.64 ($p < 10^{-13}$), 2.66 ($p < 10^{-9}$), respectively. Since CNR scales with the square root of imaging dose (Fig. 9), changing from TBL + LFB to IBL + LFB and IBL + LFB to IBL + SPA reduces the imaging dose required to obtain a given CNR by factors of $(1/1.63)^2 = 0.38$ and $(1/1.64)^2 = 0.37$, respectively. Over multiple contrast inserts and phantom sizes, the IBL + SPA system produced the...
same CNR as the TBL + LFB system at, on average, a factor of \((\frac{1}{2.66})^2 = 0.14\) of the imaging dose.

Table IV shows for both phantoms that the theoretical CNR ratios were within 5% of the measured values for the TBL + LFB to IBL + LFB imaging system configuration change, indicating good agreement between theory and measurement. The measured CNR ratios for the IBL + LFB to IBL + SPA change were 14% below (EMMA phantom) and 22% below (LEMMA phantom) the theoretical values. We expect this disagreement is due to the suboptimal optical coupling and alignment between SPA pixels and a-Si detectors, which is discussed in greater detail in Sec. IV F.

### IV.E MTF

The plots in Fig. 10 show that the mean MTFs (over all dose) for a given spatial frequency for both phantoms and all three imaging systems differed from each other by 0.06 or less at all spatial frequencies considered. The MTFs were lower for the LEMMA than the EMMA phantom in large part because of the use of the “edge enhancing” and “smoothing pelvis” reconstruction protocols, respectively. The MTF standard deviations (over all imaging doses) for a given frequency over all imaging doses never exceeded 0.05, suggesting that the MTFs are weakly dependent on imaging dose. Table III shows that the MTF percentage point differences were statistically significant at the \(p < 0.07\) level for the EMMA and LEMMA cases except for the LEMMA case with the TBL + LFB to IBL + SPA change (\(p = 0.42\)). When both phantoms were considered together, the MTF percentage point differences were statistically insignificant for all imaging system changes. All mean MTF percentage point differences between systems were within 3%, as shown in Table III. Although the MTF differences were statistically significant, this is an insignificant MTF difference from a practical standpoint. Indeed, Gopal, and Samant concluded in a rigorous statistical analysis of bar pattern versus edge detection and slit-based MTF calculation methods that the three methods differed insignificantly from each other at the 3% level.38

### IV.F Optical coupling and alignment between SPA pixels and a-Si detectors

Due to the lack of an optimized construction process, SPA pixels were not optically coupled or rigorously aligned with the a-Si detectors in the EPID. Thus, shifts of the SPA scintillator of less than 0.4 mm inside the flat panel imager between and during imaging sessions could alter the detector response, providing suboptimal IBL + SPA image quality. It is not known if or how frequently such shifts occurred during or between the image acquisition sessions that contributed to the current work. The results nevertheless demonstrate that the IBL + SPA system improves CNR relative to the
Fig. 9. Effect of imaging dose on contrast-to-noise ratio (CNR) for two phantoms and two contrast insert relative electron densities. Error bars are one standard deviation of the CNR calculated on the 3D image slices from Fig. 5. The smooth blue curves are square root functions fit to the CNR data for each imaging system.

Fig. 10. Modulation transfer functions (MTF) for (a) the head-sized phantom (EMMA) and (b) the pelvis-sized phantom (LEMMA). The lines and the error bars represent the mean and standard deviation, respectively, of the MTFs measured for all seven phantom imaging doses considered.
TABLE II. CNR ratios obtained by changing from imaging system X to imaging system Y (X→Y). CNR ratios were calculated for each imaging dose in the 3–60 cGy range; averages and standard deviations were calculated over all CNR ratios in the dose range.

<table>
<thead>
<tr>
<th>Phantom</th>
<th>rED</th>
<th>Mean</th>
<th>Std. dev.</th>
<th>n</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>EMMA</td>
<td>1.09</td>
<td>1.84</td>
<td>0.52</td>
<td>6</td>
<td>&lt;10^-2</td>
</tr>
<tr>
<td>EMMA</td>
<td>1.48</td>
<td>1.72</td>
<td>0.06</td>
<td>6</td>
<td>&lt;10^-7</td>
</tr>
<tr>
<td>LEMMA</td>
<td>1.09</td>
<td>1.60</td>
<td>0.59</td>
<td>5</td>
<td>&lt;10^-3</td>
</tr>
<tr>
<td>LEMMA</td>
<td>1.48</td>
<td>1.35</td>
<td>0.08</td>
<td>6</td>
<td>&lt;10^-4</td>
</tr>
<tr>
<td>All</td>
<td>1.63</td>
<td>0.41</td>
<td>0.04</td>
<td>23</td>
<td>&lt;10^-5</td>
</tr>
</tbody>
</table>

Notes: CNR: Contrast-to-noise ratio, EMMA: head-sized image quality phantom, LEMMA: pelvis-sized image quality phantom, rED: contrast region electron density relative to water. n: number of imaging doses considered, p: p-value from one-sided, paired, t-test (see text). TBL→IBL: Treatment beam line (TBL) equipped with Lanex Fast-B scintillator (LFB), IBL→LFB: Imaging beam line (IBL) with LFB, IBL→SPA: IBL with sintered pixelated array (SPA) scintillator.

TABLE III. Percentage point changes in mean MTF over all imaging doses obtained by changing from imaging system X to imaging system Y (X→Y).

<table>
<thead>
<tr>
<th>Phantom</th>
<th>Mean</th>
<th>Std. dev.</th>
<th>n</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>EMMA</td>
<td>1.4%</td>
<td>1.9%</td>
<td>6</td>
<td>0.07</td>
</tr>
<tr>
<td>LEMMA</td>
<td>2.7%</td>
<td>3.3%</td>
<td>6</td>
<td>0.05</td>
</tr>
<tr>
<td>All</td>
<td>0.6%</td>
<td>3.3%</td>
<td>12</td>
<td>0.26</td>
</tr>
</tbody>
</table>

Notes: MTF: Modulation transfer function, EMMA: head-sized phantom, LEMMA: pelvis-sized phantom, n: number of spatial frequencies considered, TBL→IBL: Treatment beam line (TBL) equipped with Lanex Fast-B scintillator (LFB), IBL→LFB: Imaging beam line (IBL) with LFB, IBL→SPA: IBL with SPA system.

TBL + LFB and IBL + LFB systems while maintaining a comparable MTF. It is expected that an optimized method for optical coupling and alignment of the SPA with the a-Si: H detectors would further improve the IBL + SPA results.

The SPA detector prototype is currently being improved by precision optical alignment between the SPA and the photodiode array, and by bonding the SPA directly to the a-Si:H substrate using optically-matched UV-activated adhesive. These measures are likely to improve MTF further and remove the sources of ring artifacts at the interfaces between adjacent SPA tiles.

IV.G TBL + SPA system

The IBL + LFB and IBL + SPA images were acquired with two clinical, IBL-equipped linear accelerators: one with an LFB and one with an SPA. A TBL + SPA system was not considered in this work because only one SPA was available. Switching the IBL + SPA system to a TBL + SPA system during the clinical workday would have required 1 h or more of linear accelerator down time and was considered too clinically inconvenient to be justified. The study would have been more complete had a TBL + SPA system been considered, but it is not expected that a TBL + SPA system would have provided superior images to the IBL + SPA system at a given dose.

V. CONCLUSIONS

Since CNR scales with the square root of imaging dose, changing from TBL + LFB to IBL + LFB and IBL + LFB to IBL + SPA reduces the imaging dose required to obtain a

TABLE IV. Theoretical (Theo.), measured (Meas.), and percentage difference (% Diff.) between mean measured (Table II) and expected contrast-to-noise ratio (CNR) ratios from for each imaging system configuration change and phantom.

<table>
<thead>
<tr>
<th>Phantom</th>
<th>TBL→IBL+LFB</th>
<th>IBL→IBL+SPA</th>
<th>TBL→IBL+SPA</th>
</tr>
</thead>
<tbody>
<tr>
<td>EMMA</td>
<td>1.78</td>
<td>1.78</td>
<td>0</td>
</tr>
<tr>
<td>LEMMA</td>
<td>1.40</td>
<td>1.48</td>
<td>5</td>
</tr>
</tbody>
</table>

Notes: EMMA: head-sized phantom, LEMMA: pelvis-sized phantom, TBL→LFB: Treatment beam line (TBL) equipped with Lanex Fast-B scintillator (LFB), IBL→LFB: Imaging beam line (IBL) with LFB, IBL→SPA: IBL with SPA system.

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given CNR by factors of 0.38 and 0.37, respectively. Comparable spatial resolution was observed with all imaging system configurations. IBL and SPA patient image quality was always better than that of the TBL + LFB system and as good as or better than that of the IBL + LFB system.

ACKNOWLEDGMENTS

Two of the authors (EKB, DSE) were supported by research fellowships from the Iowa Center for Research by Undergraduates (ICRU). The authors thank Kellie Bodeker, M.S., for obtaining IRB approval for the patient studies, Jane Hershberger for obtaining consent from the patients who were imaged, and Gareth Smith for preparing many of the figures. This work was partially supported by Siemens Oncology Care Systems.


