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Spectral imaging using clinical megavoltage beams and a novel multi-layer imager

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Abstract

We assess the feasibility of clinical megavoltage (MV) spectral imaging for material and bone separation with a novel multi-layer imager (MLI) prototype. The MLI provides higher detective quantum efficiency and lower noise than conventional electronic portal imagers. Simulated experiments were performed using a validated Monte Carlo model of the MLI to estimate energy absorption and energy separation between the MLI components. Material separation was evaluated experimentally using solid water and aluminum (Al), copper (Cu) and gold (Au) for 2.5 MV, 6 MV and 6 MV flattening filter free (FFF) clinical photon beams. An anthropomorphic phantom with implanted gold fiducials was utilized to further demonstrate bone/gold separation. Weighted subtraction imaging was employed for material and bone separation. The weighting factor (w) was iteratively estimated, with the optimal w value determined by minimization of the relative signal difference (S_R) and signal-difference-to-noise ratio (SDNR) between material (or bone) and the background. Energy separation between layers of the MLI was mainly the result of beam hardening between components with an average energy separation between 34 and 47 keV depending on the x-ray beam energy. The minimum average energy of the detected spectrum in the phosphor layer was 123 keV in the top layer of the MLI with the 2.5 MV beam. The w values that minimized S_R and SDNR for Al, Cu and Au were 0.89, 0.76 and 0.64 for 2.5 MV; for 6 MV FFF, w was 0.98, 0.93 and 0.77 respectively. Bone suppression in the anthropomorphic phantom resulted in improved visibility of the gold fiducials with the 2.5 MV beam. Optimization of the MLI design is required to achieve optimal separation at clinical MV beam energies.

Keywords

EPID; MLI; Spectral; Bone separation; Material separation

1. Introduction

Spectral x-ray imaging (also referred to as dual-energy imaging) was introduced by Alvarez and Macovski (1976). They described multi-parametric x-ray imaging with kilovoltage (kV) computed tomography (CT) acquisitions and subsequent estimation of the atomic number and mass density of the imaged material. As a result, imaged objects could be identified based on their composition and pseudo-monoenergetic images could be reconstructed within the range of the initial x-ray beam energy. Concurrently, an early realization of material discrimination using spectral imaging in the clinical setting was reported by Rutherford *et al* (1976). They used an EMI brain scanner (New *et al* 1974) and two x-ray beam energies to reconstruct information on material atomic number and electron density. Other techniques to perform dual-energy imaging followed; Brooks and Di Chiro (1978) used a stacked detector geometry to capture different parts of the x-ray spectrum emitted from a single source during a single acquisition, whereas Kalender *et al* (1986) utilized a fast switching x-ray tube to perform dual kV CT.

Translation of spectral imaging from the diagnostic to the radiotherapy environment has been demonstrated using the kV on-board imaging system mounted on the clinical linear accelerator (LINAC) gantry orthogonal to the MV treatment beam. Hoggarth *et al* (2013) used dual-energy to subtract bone for improved soft-tissue and tumor visualization during image-guided lung radiotherapy. In a subsequent study they used their technique for markerless motion tracking of lung tumors with dual-energy fluoroscopy (Patel *et al* 2015). The technique was based on Brody *et al* (1981) who applied dual energy imaging to separate bone and soft tissue in chest radiography using 70 and 140 kVp x-ray beams. However, there are several challenges with this technique due to the sequential, non-synchronous acquisitions. Another approach from Yang *et al* (2011), used lower energy MV beams (~1 MV) for dual-energy imaging but their study was focused on the determination of stopping-power ratios for proton therapy with higher accuracy than dual-energy kV beams.

In radiation therapy, the primary MV beam spectrum contains polychromatic x-rays that can potentially be sampled separately. Possible applications include a) improved soft-tissue contrast for patient positioning and treatment verification; b) bone “removal” to better visualize tumors in the lung or fiducials in other locations; c) real-time tumor localization and tracking; d) metal artifact reduction; and, e) adaptive treatment planning through improved soft-tissue contrast, accurate tissue identification and MV cone-beam computed tomography (MVCBCT).

The current conventional EPID design is suitable for patient setup and treatment verification but does not take advantage of the range of energies provided by the MV x-ray beam. An attractive imager architecture for MV spectral imaging is the stacked detectors arrangement initially proposed by Brooks and Di Chiro (1978) to perform dual-energy kV imaging. They used a two-layer configuration consisting of a thin, highly-absorptive CaF₂ scintillator on-top to capture low-energy x-rays and a thick NaI scintillator at the bottom to capture the rest of a 100 kVp spectrum. They demonstrated better x-ray energy separation at low energies compared to two separate exposures at different x-ray beam energies. The concept of using a top imager layer as a filter for a bottom imager layer can be readily translated to the

radiation therapy MV x-ray energy range, where fast energy switching or dual-head MV irradiation are not available.

A novel MV multi-layer imager (MLI) design with stacked detector configuration has been recently introduced (Rottmann *et al* 2016). The MLI prototype is composed of four stacked layers of copper (Cu) buildup, gadolinium oxysulfide (GOS) phosphor and an a-Si flat panel array (figure 1). Due to beam hardening from the passage of the beam from successive Cu/GOS/a-Si layers, the spectrum of the deposited energy in each GOS component at each layer is expected to be different, mimicking energy-separation between the layers. The aims of the current work are: a) to identify the MLI energy absorption characteristics that contribute to energy separation required for spectral imaging using a previously validated Monte Carlo model of the MLI (Myronakis *et al* 2017) and b) to experimentally evaluate the feasibility of MV spectral imaging using the MLI prototype for material and bone separation.

2. Methods

2.1. Multi-layer imager prototype design

The MLI design prototype is composed of four conventional EPID layers (figure 1). The main components of each layer are an 1 mm Cu buildup plate, a terbium doped GOS phosphor with thickness equal to 0.490 mm and a photodiode array. The prototype size specifications have been designed to fit in the current clinical EPID on the Varian TrueBeam platform (AS-1200). The frame-rate was 15 fps, identical to the clinical AS-1200. The design prototype demonstrated 1.7 times higher signal-to-noise ratio (SNR) and five times higher detective quantum efficiency at zero frequency DQE(0) than the AS-1200, while maintaining performance in the modulation transfer function (MTF). Image acquisition from each layer or/and combination of layers is feasible. Further details on the MLI design specifications and performance characteristics can be found in Rottmann *et al* (2016).

2.2 Monte Carlo model of MLI

The Geant4 Application for Tomographic Emission (GATE) application (Jan *et al* 2011, Sarut *et al* 2014) was utilized to develop a Monte Carlo model for the MLI (Myronakis *et al* 2017). X-ray, charged particle and optical photon interactions and transport were simulated for each layer of the MLI. The available output at the end of each simulation includes but is not limited to optical photon detection, energy absorption at the phosphor and Compton scatter from each MLI component. The model has been previously validated against measured data of MTF, Noise Power Spectrum (NPS) and DQE and demonstrated very good agreement (Myronakis *et al* 2017).

2.3. MLI beam hardening and energy separation calculations

Monte Carlo computations were performed using the validated model of the MLI described in the previous section to assess x-ray absorption, beam hardening and subsequent energy separation between MLI layers. Monte Carlo computations were performed with and without a phantom in the beam path. The phantom used was a $27 \times 27 \times 27$ cm³ box of water,

reflecting an average body thickness. The energy spectra used in the simulations were provided by Varian.

2.4. Weighted subtraction of material layers in solid water slabs

2.4.1. Experimental setup—A solid water phantom composed of two slabs, each with thickness equal to 5 cm, was used for the experimental work. Thin layers of Aluminum (Al), Cu and Gold (Au) were placed between the solid water (figure 2a). Aluminum ($Z=13$) has a similar atomic number and almost identical attenuation characteristics as bone. Gold ($Z=79$) is often used for implanted fiducials; and Cu ($Z=29$) was included as an intermediate atomic number material.

A Varian Truebeam LINAC equipped with 2.5 MV, 6 MV and 6 MV flattening filter free (FFF) beams was used for irradiation. The MLI was positioned at 150 cm source-to-detector-distance (SDD). The middle of the phantom was positioned at isocenter and the field size was 10 by 10 cm² at isocenter (figure 2b). The phantom was irradiated using 2.5 and 6 MV FFF beams to maximize the low-energy x-ray content that contributes to image contrast. Each image was generated as an average over fifty images per frame at 1.9 MU for 2.5 MV and 5.5 MU for 6 MV FFF per frame, respectively. The number of MU was heuristically determined at each beam energy to avoid image saturation. The dose rate was 60 MU/min at 2.5 MV and 1400 MU/min at 6 MV FFF. Full-field images for corrections were also acquired at the same settings. The full field images (I_F) were acquired at the same MU for each energy. The multi-leaf collimators (MLCs) were retracted and jaws were opened for the radiation field to cover the full detector surface. Each I_F image was averaged over 100 frames. The gain (G) was calculated as the average intensity in I_F . Two images were acquired at each irradiation. The first image was obtained from the top layer and the other image from the combination of the subsequent three layers. The images (I) were corrected for dark field (I_D), I_F and G according to the following formula

$$I = \frac{I - I_D}{I_F - I_D} \cdot G \quad (1)$$

2.4.2. Weighting factor determination—Skhumat *et al* (2008) described a dual-energy (DE) weighted log-subtraction imaging (WSI) scheme to decompose soft-tissue and bone in chest radiography. The same principle was applied in the current work to assess the feasibility of DE-WSI with clinically used MV beams. Two images were acquired at different layers of the MLI. One of the images was multiplied with a weighting factor (w) and then subtracted from the other image to produce a weighted subtracted image.

The optimal weighting factor value was iteratively determined by calculating the signal difference (S_R) between the background signal and the signal of each material (Hoggarth *et al* 2013)

$$\Delta S_R = (\bar{I}_m - \bar{I}_b) \quad (2)$$

$\bar{I}_m$ corresponds to the average signal in a region of interest (ROI) for each of the materials considered in this study (i.e. Al, Cu, Au, bone) and $\bar{I}_b$ corresponds to the average signal in a background ROI (i.e. solid water or soft tissue). The signal-difference-to-noise ratio (SDNR) was evaluated for the weighted subtracted image for each material using the following formula (Shkumat *et al* 2008):

$$SDNR = 2 \frac{(\bar{I}_m - \bar{I}_b)}{(\sigma_m + \sigma_b)} \quad (3)$$

σ_m and σ_b are the average noise over the ROI on each material and the background, respectively.

A square ROI of 50 by 50 pixels was used to calculate average signal and noise values. At each iteration, w was incrementally increased across a large pre-determined range. The optimal value of w for each material was defined as the value that produced zero relative signal between the background and the material in the weighted subtracted image (i.e. $S_R \approx 0$). The weighting and subtraction of the acquired images was performed on the natural logarithms of the images.

$$\ln I_w = \ln I_2 - w \cdot \ln I_1 \quad (4)$$

where I_w is the weighted subtracted image, I_2 is the image obtained from the bottom imager layer and I_1 is the image obtained from the top imager layer.

2.5. Weighted subtraction of bone in pelvis phantom

A pelvis phantom (CT Torso Phantom CTU-41) was irradiated at 2.5 MV, 6 MV and 6 MV FFF. Two gold fiducials were placed such that they would overlap with bone in both anterior-posterior (AP) and lateral projections. Fifty dark field images were averaged to achieve sufficient statistical accuracy. Full-field images were acquired at 1 MU/frame and averaged over fifty frames. The phantom was irradiated at 1 MU/frame; one frame was acquired at each beam energy. Images were corrected according to equation (1). The field size was adjusted to irradiate the whole MLI surface area, i.e. $43 \times 43 \text{ cm}^2$ at source-detector distance (SDD) equal to 150 cm. Images were obtained from the top (Layer 1) and bottom (Layer 4) layers of the MLI. ROIs were defined within the gold fiducials and in bony areas of the pelvis phantom. Since the fiducial size was in the order of few millimeters, the corresponding ROI area was small (~ 30 pixels in total). The average signal for the fiducial and bone ROI was obtained and the optimal weighting factor was determined using the method described above.

3. Results

3.1. MLI beam hardening and energy separation

In figure 3, the spectrum of the energy deposited in each GOS layer is shown for the 2.5 MV (figure 3a), 6 MV (figure 3b) and 6 MV FFF beams (figure 3c). The spectra have been normalized to the total number of events (i.e. detected energy depositions) in the top MLI layer (Layer 1). The figure demonstrates the expected difference in absolute number of detected events in each layer. Layer 2, detected approximately 31% fewer events than the top layer. Layer 3 and Layer 4 detected 48% and 62% fewer events than the top layer, respectively. Similar trends were observed for 6 MV (figure 3b) and 6 MV FFF (figure 3c). At 6 MV (figure 3b), the difference in detected events between layers was not as great as in 2.5 MV. The mean energy of the detected events spectrum (table 1) increased with layer number (i.e. lower layers were subject to higher mean energies). For each layer, the mean energy of detected events was lower for 2.5 MV than for 6 MV or 6 MV FFF beams. For all energies, the largest difference between the means was observed between the top (Layer 1) and bottom (Layer 4) layers. This difference was largest at the 6 MV FFF beam.

Table 1 corresponds to average energies obtained without the water phantom within the beam path. In table 2, the average deposited energies were obtained with the water phantom in the beam path. The overall trend was similar to the trend observed in table 1. However, the average energy was shifted towards higher values with difference range from 6% for 2.5 MV and the top layer up to 29% for 6 MV FFF beam and the bottom layer. In addition, the difference between the average energy deposited in the top layer and the other layers was significantly increased.

3.2. Weighted subtraction of material layers

The S_R and SDNR for different weighting factor values calculated for the solid water phantom with Al, Cu and Au material sheets is displayed in figure 4. S_R and SDNR values for 6 MV FFF were lower than those for 2.5 MV. Both metrics demonstrated a steeper change at 2.5 MV compared to 6 MV FFF. In addition, the weighting factor values at which the S_R for each material equaled zero were more closely spaced for 6 MV FFF than for 2.5 MV. The w values that minimized S_R for Al, Cu and Au were 0.89, 0.76 and 0.64 at 2.5 MV. The respective w values at 6 MV FFF were 0.98, 0.93 and 0.77. When S_R for a specific material was zero, this material had effectively the same average pixel intensity as the background (figure 5), i.e. the ratio between the mean signal in a ROI within the material and the same ROI in the background was equal to one.

3.3 Weighted subtraction of bone

In figure 6, the results from the weighted subtraction of bone from MV images acquired with the 2.5 MV beam are shown (figure 6(b) and (d)). Images obtained with conventional EPID are also displayed for comparison (figure 6(a) and (c)). The gold fiducials are magnified and displayed in the insets of each separate image. AP and lateral acquisitions are shown in figures 6(a)–(b) and 6(c)–(d) respectively.

Image histogram equalization was performed in the images shown in figure 6. The results are displayed in figure 7. The window level was the same in each pair of images (a–b and c–d). Pixel intensity profiles through the gold fiducials are also incorporated in figure 7. The solid line corresponds to the pixel intensity before weighted subtraction and the dashed line corresponds to pixel intensity across the gold fiducial after weighted subtraction.

4. Discussion

The MLI was designed to provide higher detective quantum efficiency and improved noise power spectrum characteristics compared with conventional EPIDs at the photon energy range utilized in radiation therapy (Rottmann *et al* 2016). The unique multi-layer design was further exploited in the current study to evaluate spectral imaging. Towards this end, computational and experimental measurements were performed to assess the feasibility of bone suppression in MV imaging and identify strategies for further improvements to the MLI design. The MLI simulations carried-out in this work provided insight on energy deposition in the GOS phosphor, beam-hardening and subsequent energy separation between the layers of the MLI.

Energy depositions greater than 1 MeV had negligible contribution to the energy spectrum in the GOS phosphor, regardless of the initial beam energy (figure 3). One reason for this was the small thickness (436 μm) relative to the high energies considered in this work. The second reason was the reduced effective density of the phosphor compound (4.59 g/cm³) relative to the theoretically expected (7.33 g/cm³). The phosphor was a compound of GOS, a binder to hold the GOS grains and air (Myronakis *et al* 2017) and had less mass density than a phosphor composed of GOS only. Subsequently, the observed energy separation between layers was mainly attributed to beam hardening between consecutive layers and components of the MLI.

Beam hardening introduced differences between the mean energy of the spectra deposited in the phosphor at each layer; these differences were maximized between the top and the bottom layer of the MLI. The maximum difference between mean deposited energies was between 34 and 47 keV depending on initial beam energy (table 1). Interestingly, the 6 MV FFF beam demonstrated the highest mean-energy separation. This can be attributed to the higher content of lower energy x-rays and the wider energy range of the 6 MV FFF beam spectrum. Despite the highest energy separation, the mean energy of all layers is high enough to cancel any effect of improved energy separation (discussed below); the mean energy per layer will be even higher when an object is in the beam path as shown in table 2. In principle, the water phantom removed part of the low energy x-rays and the top layer absorbed most of the remaining low-energy x-rays leaving the higher energies x-rays for the subsequent layers.

The estimated spectral difference for 2.5 MV, 6 MV and 6 MV FFF was at least two times greater than for clinical dual-energy kV imaging, where the difference between the mean energies of 80 and 140 kV beams is around 15 to 20 keV*. Additional filtration in the high-energy kV beam may increase the difference. However, in MLI MV imaging the mean energy of the deposited spectra is above 120 keV (table 1); this is not ideal for bone/soft-

tissue separation considering the similar attenuation characteristics of those tissues for energies above 100 keV. In addition, an object placed in the path of the beam (e.g. phantom or tissue) will further harden the beam, increasing the mean energy of the detected spectrum well above 120 keV for the 2.5 MV beam.

Our experiments demonstrated that material removal is feasible with the current MLI configuration, which was not originally designed for this purpose. Results from figures 4 and 5 suggest that a steep change in SDNR and S_R curves combined with large spacing between the weighting factors at which SDNR and S_R are minimized can produce better material separation. This was particularly evident at 2.5 MV (figure 4(a) and (b)). However, when Au was removed, Cu and Al could also not be distinguished in the weighted subtracted image at 2.5 MV (figure 5(c)). The separation at 6 MV FFF was not very good as shown in (figure 5(d)–(f)) due to the low contrast range available in the weighted subtracted image as a result of the higher energy of the beam and the additional beam hardening from the imaged object.

The higher content of low energy x-rays at 2.5 MV along with density differences between materials can explain the weighted subtraction observed at this beam energy. However, the signal ratios calculated in figure 5 imply that the available contrast range in the weighted subtracted images was rather narrow as a result of the converging mass-attenuation coefficients at higher energies. Most important, though, was the observation that when Al contrast is minimized (figure 5(a) and 5(d)), the contrast of Au is still apparent. This implies that even with the un-optimized MLI configuration, bone suppression while maintaining fiducial contrast is possible.

This conclusion was further demonstrated using a pelvis phantom and gold fiducials. The fiducials were still visible after weighted subtraction, despite the increased noise, apparent in the weighted subtracted images. Increased pixel noise is also shown in the pixel intensity values in figure 7 obtained from fiducial cross sections. More interestingly, the visibility of the fiducials compared to their immediate vicinity improved, as shown in figures 6 and 7 even after histogram equalization, for both AP and lateral acquisitions. In MVCBCT, the low-dose per frame image acquisition imposes a limit in the available image contrast. The use of weighted subtraction to selectively enhance regions of the MVCBCT image, in combination with an optimized multilayer-design, suitable for MV spectral imaging, may overcome the contrast limitations of conventional MVCBCT.

5. Conclusion

Material separation with clinical MV beams was shown to be best at 2.5 MV using a prototype multi-layer imager consisting of four conventional GOS/a-Si layers. The spectral separation available with this detector configuration for 6 MV and 6 MV-FFF beams is not likely to be sufficient for clinical applications. This work suggests that optimization of MLI construction (e.g. scintillator material/thickness and converter layer material/thickness)

*X-ray spectra were obtained from the Siemens online spectrum calculator (<https://www.oem-xray-components.siemens.com/x-ray-spectra-simulation>). The filtration used at 80 kV was 2.5 mm Al and 5 mm Al at 140 kV.

could improve spectral separation further, potentially opening new avenues for beam's-eye-view imaging in radiation therapy.

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Figure 1.
Multi-layer (MLI) prototype design configuration (not to scale)

Figure 2.

(a) Irradiation geometry. The MLI was positioned 150 cm from source. The solid water phantom with the material layers was placed at isocenter (100 cm). (b) The placement of the material layers between two solid water slabs of equal thickness is shown from the beam's-eye-view.

Figure 3.

Deposited energy spectra in the GOS phosphor for each MLI layer at 2.5 MV (a), 6 MV (b) and 6 MV FFF (c). Spectra were normalized to the total detected energy depositions in the top layer.

Figure 4.

S_R for Al (blue), Cu (red) and Au (yellow) for different weighting factor values at for 2.5 MV (a) and 6 MV FFF (c). SDNR values for each material are shown in (b) and (d) for 2.5 MV and 6 MV FFF respectively. Please note the y-axis scale difference between the two beam energies.

Figure 5. Material separation at: (a)–(c) 2.5 MV and (d)–(e) 6 MV FFF beam energy. The ratio between the mean signal in different materials and the background is also displayed.

Figure 6. Conventional EPID images (a), (c) and weighted subtracted images (b) and (d) at 2.5 MV beam for (a)–(b): AP acquisition and (c)–(d) lateral acquisition. Magnified images of the fiducials are shown in the insets.

Figure 7.

The images are the same as in figure 6 but were processed to equalize intensity histograms and window level. The pixel intensity profile across the fiducials is also displayed. The solid line represents the pixel intensity before weighted subtraction and the dashed line is the pixel intensity profile after subtraction for the corresponding fiducial.

Mean detected energy in each layer at 2.5 MV, 6 MV and 6 MV FFF. The difference between the mean energy in Layer 1 and the mean energy in subsequent layers is also displayed.

Table 1

Beam Energy (MV)	Mean Detected Energy (MeV)				Difference with Layer 1 (MeV)			
	Layer 1	Layer 2	Layer 3	Layer 4	Layer 2	Layer 3	Layer 4	
2.5	0.125	0.140	0.151	0.163	0.015	0.026	0.038	
6	0.218	0.223	0.230	0.252	0.005	0.012	0.034	
6 FFF	0.190	0.203	0.213	0.237	0.013	0.023	0.047	

Mean detected energy in each layer at 2.5 MV, 6 MV and 6 MV FFF with water phantom in the beam path. The difference between the mean energy in Layer 1 and the mean energy in subsequent layers is also displayed.

Table 2

Beam Energy (MV)	Mean Detected Energy (MeV)				Difference with Layer 1 (MeV)			
	Layer 1	Layer 2	Layer 3	Layer 4	Layer 2	Layer 3	Layer 4	Layer 4
2.5	0.133	0.162	0.177	0.193	0.029	0.044	0.060	
6	0.226	0.258	0.273	0.307	0.032	0.047	0.081	
6 FFF	0.219	0.253	0.271	0.305	0.034	0.052	0.086	