Compton scatter imaging: A promising modality for image guidance in lung stereotactic body radiation therapy

Gage Redler^{a)}, Kevin C. Jones, Alistair Templeton, Damian Bernard, Julius Turian, and James C. H. Chu

Department of Radiation Oncology, Rush University Medical Center, Chicago IL 60612, USA

(Received 18 August 2017; revised 29 December 2017; accepted for publication 2 January 2018; published 7 February 2018)

Purpose: Lung stereotactic body radiation therapy (SBRT) requires delivering large radiation doses with millimeter accuracy, making image guidance essential. An approach to forming images of patient anatomy from Compton-scattered photons during lung SBRT is presented.

Methods: To investigate the potential of scatter imaging, a pinhole collimator and flat-panel detector are used for spatial localization and detection of photons scattered during external beam therapy using lung SBRT treatment conditions (6 MV FFF beam). MCNP Monte Carlo software is used to develop a model to simulate scatter images. This model is validated by comparing experimental and simulated phantom images. Patient scatter images are then simulated from 4DCT data.

Results: Experimental lung tumor phantom images have sufficient contrast-to-noise to visualize the tumor with as few as 10 MU (0.5 s temporal resolution). The relative signal intensity from objects of different composition as well as lung tumor contrast for simulated phantom images agree quantitatively with experimental images, thus validating the Monte Carlo model. Scatter images are shown to display high contrast between different materials (lung, water, bone). Simulated patient images show superior (~double) tumor contrast compared to MV transmission images.

Conclusions: Compton scatter imaging is a promising modality for directly imaging patient anatomy during treatment without additional radiation, and it has the potential to complement existing technologies and aid tumor tracking and lung SBRT image guidance. © 2018 American Association of Physicists in Medicine [https://doi.org/10.1002/mp.12755]

Key words: IGRT, lung SBRT, Monte Carlo, scatter imaging

1. INTRODUCTION

In most contexts, scatter is considered an unavoidable and unwanted component in radiology and radiation oncology. However, scattered radiation carries useful information because the scattering process depends on the scattering medium. For example, scattered photons have been used to measure tissue electron density *in vitro*,^{1–3} lung density and breathing function,^{4–7} and bone density/composition.^{8–12}

Forming images from Compton-scattered photons (scatter imaging) was first discussed as a diagnostic imaging modality in 1959 as a method for measuring 3D patient electron density using an ¹⁹²Ir pencil beam and focused collimators.¹³ Similar approaches were implemented using a 5.6 MeV linear accelerator¹⁴ and a ⁶⁰Co source.^{15,16} A more rapid line-by-line approach using a monoenergetic ¹³⁷Cs source and energy-based spatial localization was also developed.^{17,18} Even following the widespread adoption of CT, additional Compton scatter imaging methods were developed and commercially produced.^{19–21} Compton scatter tomography,²² Compton scatter radiography for 3D mammography,²³ and coded aperture x-ray scatter imaging^{24,25} have also been proposed. Despite the technical advances, however, with the advent of CT most interest in using scatter imaging modalities for 3D electron density measurements waned due to dose limitations in noise and resolution.^{26–28}

Scatter imaging still presents some potential advantages, including the fact that it provides a more direct measure of electron density (useful for radiation therapy applications), theoretical contrast advantages when imaging small objects with large density differences, and increased imaging geometry flexibility. The main limitation is the high dose necessary to obtain scatter signal of adequate resolution and signal-to-noise ratio (SNR).

One scenario in which scatter imaging may prove useful is lung stereotactic body radiation therapy (SBRT), where large radiation doses are delivered rapidly to a region containing a small object (tumor) with increased density compared to surrounding tissues (lung). Lung SBRT is an effective treatment for early-stage nonsmall cell lung cancer when comorbidities preclude anatomical resection.^{29,30} With an aging population and increased screening, early-stage lung cancer patients are increasing in number,³¹ increasing the importance of lung SBRT.³² Lung SBRT efficacy depends on the ability to deliver high doses with submillimeter accuracy, making knowledge of target location and surrounding anatomy crucial. However, lung tumors may move centimeters due to breathing motion.³³ Therefore, image guidance in lung SBRT is paramount to successful implementation.

There currently exist a number of modalities for image-guided radiation therapy (IGRT), each of which has

advantages and disadvantages.³⁴ Recently, a related scatter imaging scheme for real-time lung tumor tracking during radiotherapy was proposed, utilizing a slit kilovoltage x-ray beam to generate scattered photons.³⁵ We have devised an approach to forming images from photons scattered during treatment as an additional tool for IGRT in lung SBRT. Unlike previous scatter imaging methods, which actively irradiated patients for collecting scatter, the proposed technique passively collects photons unavoidably generated during radiation therapy, obviating imaging dose concerns. The potential of scatter imaging is explored via experimental phantom images as well as Monte Carlo simulations of phantom and lung SBRT patient images.

2. MATERIALS AND METHODS

2.A. Basis of scatter image formation

During lung SBRT, a large amount of external beam radiation is delivered to a region containing the tumor. Some interacts and deposits energy locally, while some is carried away as scatter. A pinhole collimator localizes the origin of detected scatter, thus forming a scatter image (Fig. 1).

From a source photon's perspective using a ray-tracing formalism, the probability (P_{ijk}) of starting at point i (the source with energy $h\nu$), traveling through material to voxel j , scattering within voxel j (toward detector element k), and then (at a reduced energy $h\nu'$) reaching k is given by the product of three factors:

$$P_{ijk} = \left[\exp \left(- \sum_n \bar{\mu}_n l_n \right) \right]_{ij} \times \left[\rho_{e,j} \frac{d_e \sigma}{d\Omega_{\theta_{jk}}} d\Omega_{\theta_{jk}} l_j \right] \times \left[\exp \left(- \sum_m \bar{\mu}'_m l'_m \right) \right]_{jk}, \quad (1)$$

FIG. 1. Scatter images are formed by photons scattered from the SBRT treatment field, collimated by a pinhole collimator and collected by a flat-panel detector. [Color figure can be viewed at wileyonlinelibrary.com]

where space has been discretized into voxels, as diagrammed in Fig. 2.

The first and third bracketed factors account for exponential attenuation and represent the probability that the photon is transmitted from points i to j and j to k , respectively. While traveling from i to j (or j to k), the photon passes through n (or m) voxels, with a voxel-specific interaction length (l_n or l'_m , units of cm) and average total attenuation coefficient ($\bar{\mu}_n$ or $\bar{\mu}'_m$, units of cm^{-1}).

The central bracketed factor represents the probability that the photon entering voxel j is scattered into the solid angle subtended by detector element k ($d\Omega_{jk}$, units of sr). As with $d\Omega_{jk}$, the scattering probability is also linearly proportional to the electron density of voxel j ($\rho_{e,j}$, units of electrons per cm^3), the interaction length of the source ray within voxel j (l_j , units of cm), and the energy- and angle-dependent differential Compton cross section ($\frac{d_e \sigma}{d\Omega_{\theta_{jk}}}$, units of $\text{cm}^2 \text{sr}^{-1}$ per electron). Thus, the higher the electron density and the longer the photon ray interaction length within voxel j , the higher the scatter probability.

Equation (1) represents the probability of a photon traveling between three points (i to j and j to k , with scattering angle θ_{jk}) via a single scattering event. In reality, multiple scattering also occurs (and could be described by adding additional scattering and exponential attenuation factors to Eq. (1), similar to the existing second and third factors in Eq. (1)). Although the presented work does not explicitly use

FIG. 2. Diagram of the processes behind proposed scatter image formation (simulated using Monte Carlo) including attenuation of incident radiation (from source located at point i) with energy $h\nu$ before reaching scattering element v_j with some probability of scatter event along angle θ_{jk} connecting v_j and detector element d_k as well as attenuation of scattered radiation with energy $h\nu'$ exiting the patient toward d_k . Corresponding mathematical description is in Equation (1). [Color figure can be viewed at wileyonlinelibrary.com]

Eq. (1) for its simulations (instead, Monte Carlo is used), the above discussion and equation are intended to illustrate the important processes that must be considered to understand the approach to scatter imaging discussed in this work. Rather than rigorously model all facets of the radiation transport physics determining scatter image formation (as can be done with Monte Carlo), Eq. (1) describes, for any given source photon, the dominating first-order processes determining its probability of contributing to the detected scatter image. Specifically, this equation is meant to emphasize that: (a) scatter image intensity will be affected by both entrance and exit attenuation; (b) the probability of Compton scattering is proportional to electron density (which is roughly proportional to physical density), and therefore, that scatter imaging contrast will originate in these density differences; and (c) scatter image intensity will be dependent on the angular orientation between source, object, and detector apparatus. A more rigorous characterization of the scatter imaging discussed in this work, using an analytical simulation method, is the subject of a separate work.³⁶

2.B. Experimental imaging

Scatter images were acquired during phantom irradiation with a Varian TrueBeam linear accelerator (linac) (Varian Medical Systems, Palo Alto, CA, USA), calibrated to deliver 1 cGy/MU at isocenter and $d = 1.5$ cm with a 10×10 cm² field in water. Parameters approximating those of clinical lung SBRT treatment were used: 6 MV flattening-filter-free (FFF) beam and 1200 MU/min dose rate. Jaw-based collimation was used to define the fields.

The scatter imager consisted of a pinhole collimator and flat-panel detector. The collimator, previously used for ^{99m}Tc nuclear medicine studies, had a 0.5 cm diameter pinhole. The conical collimator had a 25.0 cm diameter field-of-view (FOV), spaced 18.5 cm from the pinhole. The collimator thickness was equivalent to 7.0 mm Pb. The flat-panel detector was a Konica Minolta RP-4S110 17" $\times$ 14" computed radiography (CR) cassette (Konica Minolta Medical & Graphic, Inc., Tokyo, Japan) with 175 μ m detector pitch. Raw images were resampled by a factor of 10 to increase SNR, resulting in 0.17×0.17 cm² pixels.

To investigate scatter image contrast and ability to differentiate relevant tissues, three cylinders ($d = 2.8$ cm, $h = 7.0$ cm) of water ($\rho = 1.00$ g/cm³), bone ($\rho = 1.824$ g/cm³), and lung ($\rho = 0.29$ g/cm³) equivalence were imaged. The treatment beam encompassed the cylinders, and images were acquired from radiation scattered at $\sim 90^\circ$ from the beam central axis with 1:1 magnification at isocenter (Fig. 3).

Figure 3 also shows a simplified lung tumor phantom (LTphan) constructed from bolus material (representing soft tissue; Vanarsdale Innovative Products Inc., Pensacola, FL, USA), cork (representing lung tissue), and Virtual WaterTM (representing tumor tissue; Med-Cal, Inc, Verona, WI, USA). The LTphan was positioned so that the treatment beam passed through 4 cm of bolus (approximating chest wall), followed by 3.1 cm of cork, before impinging upon the

FIG. 3. Geometry used for experimental and simulated phantom scatter images. Treatment beam and scatter imager orientation are shown for both the three cylinder phantom and LTphan. For the three cylinder phantom, the materials are (in order coming out of the page) lung (orange), water (blue), and bone (gray) equivalent. All cylinder dimensions are $d = 2.8$ cm and $h = 7.0$ cm. For the LTphan, the treatment beam passes from left to right through 4 cm water before reaching a $9 \times 9 \times 9$ cm³ region of lung with a Virtual WaterTM cylinder embedded at its center. There are 2 cm of water on the top/bottom of the LTphan. For both phantom setups, the beam axis intersects the pinhole collimator axis at the beam isocenter, which is set at the center of the water cylinder. [Color figure can be viewed at wileyonlinelibrary.com]

Virtual WaterTM cylinder tumor target ($d = 2.8$ cm, $h = 7.0$ cm, long axis perpendicular to beam central axis). The scatter imager acquired an image from radiation scattered at $\sim 90^\circ$ from the beam central axis, with 1:1 magnification at isocenter. Thus, the scatter was attenuated by at least 2 cm of cork and 2 cm of bolus before exiting the phantom. The 8×8 cm² treatment field encompassed the tumor target (centered at isocenter).

Experimental images were acquired with various delivered doses: as high as 5000 MU (approaching ideal noiseless images) down to 10 MU (representing clinically relevant dose and temporal resolution) or equivalently 250 s down to 0.5 s temporal resolution. Since the collimator was designed to attenuate lower energy radiation (~ 140 keV), high-energy scatter and high-energy leakage radiation from the adjacent linac treatment head were not appropriately attenuated, and created broad background artifacts. These artifacts were suppressed by subtracting background images (with no object) from the phantom images.

Signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR) quantitatively describe the image quality as a function of imaging dose (or time). Noise magnitude (N) is defined as the standard deviation in a local uniform region of the image. SNR is defined as $\bar{T}/N$, where $\bar{T}$ is average tumor signal. CNR is defined as $(\bar{T} - \bar{L})/N$, where $\bar{L}$ is the average signal from the surrounding lung.

2.C. Simulation model

MCNP6 v1.0 (Los Alamos National Security, LLC, Los Alamos, NM, USA) was used for Monte Carlo simulations³⁷ and was chosen for its general flexibility and capability to simulate the nonstandard scatter imaging problem geometry. Scatter images were simulated using the built-in radiography tally (FIR:P) that generates a user-defined grid of point detectors and increases simulation efficiency with deterministic calculations of detection probability at each point every time an interaction occurs during the standard Monte Carlo random walk.^{38,39} This is less random in nature due to the pseudo-Monte Carlo approach. Additionally, this was used as a photon-counting tally and ignored energy-dependent detector efficiency. The tally was constructed to approximate the rectangular portion of the experimental CR cassette encompassing the circular FOV defined by the collimator, downsampled by a factor of 10 ($25.16 \times 25.16 \text{ cm}^2$, $148 \times 148 \text{ pixels}^2$). A truncated cone with the dimensions of the experimental collimator described above was used to exclude scatter not passing through the pinhole (ideal collimation). The treatment source was approximated as an isotropic point source with energy spectrum defined using the experimental clinical beam spectrum. The point source was 100 cm from isocenter, mimicking the linac geometry. Four exclusion planes provided arbitrary rectangular collimation, simulating asymmetric linac jaws. Object- and imager-related MCNP surfaces and cells (i.e., excluding source-related components) rotated arbitrarily about the isocenter, simulating gantry rotation.

2.D. Digital phantoms

Material atomic compositions for Monte Carlo simulations were drawn from the PNNL compendium of material compositions.⁴⁰ Materials of interest were air ($\rho = 1.205 \times 10^{-3} \text{ g/cm}^3$), water ($\rho = 1.000 \text{ g/cm}^3$), lung ($\rho = 0.290 \text{ g/cm}^3$), and bone ($\rho = 1.824 \text{ g/cm}^3$), with water approximating undefined soft tissue. The above density values and materials were used as constants for the digital phantoms rather than varying them as a function of temperature or more specific chemical compositions.

For validation of the simulation model and direct comparison to experimental images, digital representations of the experimental phantoms were created (Fig. 3). To generate scatter images of these digital phantoms, $\sim 9 \times 10^6$ source particles were simulated, providing average scatter image pixel intensity uncertainties of 2.8% for the three cylinders and 1.5% for the LTphan.

Simulated scatter images of a lung SBRT patient were generated using 4DCT data as the basis of a digital phantom. CT images were acquired with the Brilliance CT Big Bore (140 kVp, helical scan) (Philips Healthcare, Cleveland, OH, USA). Each CT slice (512×512 voxels of size $0.1172 \times 0.1172 \times 0.2000 \text{ cm}^3$) was downsampled by a factor of 4 to increase program efficiency. Slice thickness remained 2 mm, but only 41 slices (8.2 cm slab in superior–inferior direction) were used for the digital phantom. The simulated isocenter was at the approximate centroid of the tumor across the breathing cycle. Planes perpendicular to either the x, y, or z axes created surfaces for voxel boundaries in MCNP to define each CT voxel as an MCNP cell. Physical density for each voxel was independently defined using the CT Hounsfield unit (HU) to density lookup table in routine clinical use for the specific CT scanner. Voxels were assigned material type based on empirically determined HU ranges (air: $\text{HU} < -940$, lung: $-940 < \text{HU} < -200$, water/tissue: $-200 < \text{HU} < 120$, bone: $\text{HU} > 120$). Depending on simulated treatment field size, 1.4×10^6 – 1.8×10^7 source particle histories generated patient scatter images with average pixel intensity statistical uncertainties of $< 3\%$.

3. RESULTS

3.A. Experimental phantom images

The LTphan (Fig. 3) was imaged to investigate potential utility of scatter images (Fig. 4). Using a fixed 1200 MU/min dose rate, delivered dose is proportional to imaging time. Long acquisitions (5000 MU or 250 s) demonstrate tumor contrast in a low-noise image. As dose or time decreases, image quality (SNR and CNR) decreases. However, short, more clinically relevant, acquisitions (10 MU or 0.5 s) provide sufficient image quality to discern tumor (the central circular image feature) among surrounding lung. The entrance chest wall (high-intensity rectangular feature at top of image) and mediastinum (rectangle at bottom of image) are also identifiable. The quality of the 10 MU image suggests that scatter images could approach real-time feedback on patient anatomy during treatment.

3.B. Monte Carlo model validation

To validate the simulation model against the current experimental methodology, simulated and experimental phantom images were compared (Fig. 5). The simulation model ignores certain characteristics present experimentally (e.g., noise, imperfect collimation, detector efficiency), so the comparison is based on relative image intensity for varying geometries/materials. Listed uncertainties are coefficients of variation within the ROIs. Experimental images with extremely high-delivered doses (5000 MU) and imaging time (250 s) are used to suppress the effects of noise present in the experimental images but not in the simulated images.

Experimental (5000 MU) and simulated images of the three cylinders (bone, water, lung equivalent) are shown in

FIG. 4. Experimental LTphan images acquired with varying imaging dose/time and corresponding image quality metrics.

FIG. 5. (a) Experimental 5000 MU and (b) simulated images of the three cylinders (left to right: bone, water, lung). Beam direction (with respect to phantom) is from top to bottom for the cylinder phantom images. (c) Experimental 5000 MU and (d) simulated images of the LTphan. Beam direction is from left to right for LTphan images.

Figs. 5(a) and 5(b), respectively. Average intensity (relative to water) for simulated cylinder images match those of the experimental cylinder images, within the uncertainty (see Table I). This validates that the model properly simulates scatter image formation and reproduces experimentally observed features. Furthermore, these images demonstrate the ability to provide contrast between different materials relevant to lung SBRT treatment/imaging.

Experimental (5000 MU) and simulated images of the LTphan are in Figs. 5(c) and 5(d), respectively. To compare the images, the tumor contrast was calculated: (average tumor signal–average lung signal)/(average tumor signal). As shown in Table I, simulated tumor contrast matches experimental

tumor contrast, further validating the Monte Carlo model under conditions approximating those of lung SBRT.

3.C. Simulated patient images

The experimentally validated Monte Carlo scatter image simulation method was applied to lung SBRT patient 4DCT data to simulate scatter images for different breathing phases. The top row of Fig. 6 shows representative coronal slices from the CT for full inhalation and full exhalation breathing phases. The second row of Fig. 6 shows simulated scatter images using rectangular treatment fields for full inhalation and full exhalation. In the anterior–posterior (A–P) direction, the field was 5 cm wide at isocenter, essentially defining the amount of over/underlying tissue irradiated and contributing to the image signal. In the superior–inferior (S–I) direction, the field was 8.2 cm at isocenter, which encompassed the portion of the CT used for the digital phantom. The third row of Fig. 6 shows digitally reconstructed radiographs (DRRs) of the MV transmission image for full inhalation and full exhalation generated using custom-written ray-tracing code. The bottom row of Fig. 6 shows simulated kV transmission images (140 kVp) acquired orthogonal (PA) to the treatment beam axis (i.e., imaging similar anatomy and orientation to

TABLE I. Quantitative comparison of experimental and simulated phantom scatter images. Image signal intensity (relative to water) for different materials is shown for the three cylinder phantom. Tumor contrast is shown for the LTphan.

	Bone intensity	Water intensity	Lung intensity	Tumor contrast
Experimental	1.28 ± 0.08	1.00 ± 0.08	0.30 ± 0.08	0.52 ± 0.03
Simulated	1.30 ± 0.08	1.00 ± 0.08	0.30 ± 0.02	0.57 ± 0.04

FIG. 6. The top row shows coronal slices from the 4DCT data. The beam direction and irradiated tissue contributing to the scatter images are indicated. The second row shows simulated patient scatter images from these data. For comparison, the third row shows simulated kV transmission images and the bottom row shows MV transmission images orthogonal to the treatment beam axis. The lung tumor is indicated by the yellow arrow and/or red outline. Anatomic orientation is depicted by the inset cartoon patient in the bottom right corner of the images. The scatter images reproduce anatomy changes in CT due to breathing motion and have better tumor contrast relative to transmission images. (Note: MV transmission images have different anatomic orientation from other images). [Color figure can be viewed at wileyonlinelibrary.com]

the scatter images), calculated using the same ray-tracing code.

The amount of over/underlying tissue irradiated with the 5 cm A–P field size results in tumor contrast of 0.52 ± 0.08 in the scatter images. This demonstrates that radiation scattered from the patient during treatment can potentially produce images with contrast sufficient to discriminate between tumor and surrounding lung. Furthermore, the MV and kV transmission image tumor contrast were 0.26 and 0.11, respectively. This illustrates the potential for enhanced tumor contrast when imaging lung tumors using scatter vs transmission imaging.

4. DISCUSSION

Scatter imaging does not require additional radiation dose and images patient anatomy directly. Furthermore, because scattered radiation is emitted in all directions, detector placement is relatively flexible, allowing for multiple detectors to acquire 3D information. Furthermore, due to the rapidly delivered high doses for lung SBRT, phantom experiments suggest that subsecond scatter images are possible, allowing for real-time guidance (Fig. 4). Figure 6 also demonstrates that, specifically for lung tumors, scatter imaging can provide superior contrast to traditional MV or kV transmission imaging. The scatter imaging presented herein is not proposed as a replacement for current MV or kV transmission imaging, but rather as an additional, readily available source of information that can be simultaneously acquired with minimal imaging equipment and no extra dose to the patient.

While tumor contrast is comparable or superior to current clinical imaging technologies in radiation oncology, the main limitation for scatter imaging is the tradeoff between SNR or CNR and spatiotemporal resolution. Phantom images (Fig. 4) show that for temporal resolution of 0.5 s (10 MU), $\text{SNR} = 8.9$, which is potentially sufficient for tumor localization. A typical SBRT fraction involves delivery of at least 8 Gy to the tumor, which corresponds to greater than 800 MU or 80 images with 0.5 s integration. Under current experimental conditions, spatial resolution is ~ 0.5 cm and is largely dictated by pinhole size. A smaller pinhole could improve this, but SNR would subsequently decrease. Alternatively, the pinhole could be moved closer to the object to increase solid angle and thus the amount of detected scatter.

Understanding spatiotemporal resolution tradeoffs and assessing SNR in more practical scenarios will require the acquisition of clinical patient images. While the presented experimental phantom study approximates materials and geometries in lung SBRT treatments, the artificial nature of such phantom studies may bias the imaging results. For example, spheres rather than cylinders are more representative of lung tumors, but cylindrical objects were readily available and easily fabricated. Furthermore, density variations in lung tend to be smooth and not stepwise, as in the phantoms. The images in this work are of static (digital and experimental) phantoms. As an intermediate step between the presented, static work and clinical patient images, dynamic phantom studies may aid in the development of tumor motion-tracking algorithms and lead to a better characterization of the spatiotemporal resolution limitations.

Simulated patient images are generated using a Monte Carlo simulation model validated by phantom images. Certain simplifications are made that decrease model accuracy. The simulated treatment source is an isotropic point source rather than a finite, forward-peaked source. The simulated beam is perfectly collimated, thus removing collimator scatter, geometric penumbra, and edge transmission. The pinhole collimator ignores punch through, edge effects, and interactions within the collimator. Treatment head leakage is unaccounted for. The MCNP radiography tally is a photon counter and detection efficiency is energy independent. This is illustrated in Fig. 5, where signal from the beam entrance region is accentuated in the simulated vs experimental image because scatter from this region is forward scattered and therefore higher energy. Increased energy decreases experimental detector efficiency, but simulated images with perfect efficiency do not include this effect.

It is important to note that the utilized experimental equipment is suboptimal for this application. The pinhole collimator and CR cassette were designed for lower energy photons, thus decreasing the ability to collimate unwanted signal and efficiently detect useful signal. There is potential for improvements in scatter image quality via improvements in imaging equipment. Alternative collimation techniques for spatial localization can be investigated (e.g., parallel hole) as the relatively large pinhole collimator apparatus does not make optimal use of limited treatment space. Different detector approaches/technologies (electronically collimated gamma

cameras,^{41,42} photon-counting detectors,⁴³ coded aperture imaging,^{24,25} or Compton electron generating copper plates, as in EPIDs⁴⁴) could vastly improve image quality.

Treatment field size/shape dictates irradiated tissue volume, and only irradiated tissue contributes to scatter image signal. Therefore, image quality and contrast depend on field aperture. Decreasing field size in the plane parallel to the detector decreases FOV. Decreasing field size in the plane perpendicular to the detector decreases shadowing from over/underlying tissue and improves image contrast. For lung SBRT, conformal beams with decreased margins are expected to improve scatter image contrast. Future work will quantitatively evaluate the effect of field size on scatter image quality. Decreased margins will also decrease the in-plane FOV as less tissue surrounding the target is irradiated and imaged. This may affect target visualization due to a decrease in surrounding lung tissue to provide contrast in the image and may be a consideration if scatter imaging is used to track the target and/or gate the treatment beam. This would not be an issue in the cross-plane direction, which will still contain the entirety of the beam path. If intensity modulation rather than conformal therapy is used, at any given moment, only the tissue irradiated by small beamlets is imaged, potentially reducing scatter imaging applicability. This may be important as the role of intensity-modulated radiation therapy (IMRT) and volumetric modulated arc therapy (VMAT) continues to grow in modern approaches to SBRT.

5. CONCLUSIONS

In the fields of radiology and radiation oncology, scattered radiation is generally viewed as a detriment. However, scatter is inherent when using ionizing radiation, and it can provide an additional source of information. This work presents a proof of principle demonstration for Compton scatter imaging as an additional IGRT modality. This methodology provides direct images of patient anatomy during external beam radiation therapy without additional radiation dose by simply collecting inherent scattered radiation. Scatter imaging represents an interesting IGRT tool as it has potential for providing real-time, 3D images with sufficient contrast for tracking the tumor target and effectively guiding lung SBRT.

ACKNOWLEDGMENTS

This work is funded by the Swim Across America/RTSC Pilot Project Grant.

CONFLICTS OF INTEREST

The authors declare that they do not have any conflicts of interest.

^{a)}Author to whom correspondence should be addressed. Electronic mail: gage@uchicago.edu; Telephone: (773)-702-6873.

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