Implementation of radiochromic film dosimetry protocol for volumetric dose assessments to various organs during diagnostic CT procedures

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Purpose: The authors present a means to measure high-resolution, two-dimensional organ dose distributions in an anthropomorphic phantom of heterogeneous tissue composition using XRQA radiochromic film. Dose distributions are presented for the lungs, liver, and kidneys to demonstrate the organ volume dosimetry technique. XRQA film response accuracy was validated using thermoluminescent dosimeters (TLDs).

Methods: XRQA film and TLDs were first exposed at the center of two CTDI head phantoms placed end-to-end, allowing for a simple cylindrical phantom of uniform scatter material for verification of film response accuracy and sensitivity in a computed tomography (CT) exposure geometry; the TLD and film dosimeters were exposed separately. In a similar manner, TLDs and films were placed between cross-sectional slabs of a 5 yr old anthropomorphic phantom's thorax and abdomen regions. The anthropomorphic phantom was used to emulate real pediatric patient geometry and scatter conditions. The phantom consisted of five different tissue types manufactured to attenuate the x-ray beam within 1%–3% of normal tissues at CT beam energies. Software was written to individually calibrate TLD and film dosimeter responses for different tissue attenuation factors, to spatially register dosimeters, and to extract dose responses from film for TLD comparison. TLDs were compared to film regions of interest extracted at spatial locations corresponding to the TLD locations.

Results: For the CTDI phantom exposure, the film and TLDs measured an average difference in dose response of 45% ($SD \pm 2\%$). Similar comparisons within the anthropomorphic phantom also indicated a consistent difference, tracking along the low and high dose regions, for the lung (28%) ($SD \pm 8\%$) and liver and kidneys (15%) ($SD \pm 4\%$). The difference between the measured film and TLD dose values was due to the lower response sensitivity of the film that arose when the film was oriented with its large surface area parallel to the main axis of the CT beam. The consistency in dose response difference allowed for a tissue specific correction to be applied. Once corrected, the average film response agreed to better than 3% ($SD \pm 2\%$) for the CTDI scans, and for the anthropomorphic phantom scans: 3% ($SD \pm 3\%$) for the lungs, 5% ($SD \pm 3\%$) for the liver, and 4% ($SD \pm 3\%$) for the kidneys. Additionally, XRQA film measured a heterogeneous dose distribution within the organ volumes. The extent of the dose distribution heterogeneity was not measurable with the TLDs due to the limitation on the number of TLDs loadable in the regions of the phantom organs. In this regard, XRQA film demonstrated an advantage over the TLD method by discovering a 15% greater maximum dose to lung in a region unmeasured by TLDs.

Conclusions: The films demonstrated a lower sensitivity to absorbed dose measurements due to the geometric inefficiency of measuring dose from a beam situated end-on to the film. Once corrected, the film demonstrated equivalent dose measurement accuracy as TLD detectors with the added

advantage of relatively simple measurement of high-resolution dose distributions throughout organ volumes. © 2010 American Association of Physicists in Medicine. [DOI: [10.1118/1.3476455](https://doi.org/10.1118/1.3476455)]

Key words: GafChromic XRQA, Radiochromic film, CT dosimetry, TLD dosimetry, two-dimensional dosimetry

I. INTRODUCTION

As of 2006, the number of computed tomography (CT) examinations is reported to have increased by 10% per year over the past 10–15 years, and now accounts for 49% of the total collective dose (person-Sv) to patients undergoing a radiological examination; this corresponds to a 120-fold increase in CT collective dose since the 1980s.¹ Accurate patient radiation dose measurements are required to help define the significance of the increasing number of CT examinations especially within pediatric patient populations.

Since the 1980s, the industry standard technique for scanner output determination has been the computed tomography dose index (CTDI).^{2,3} CTDI was developed for single slice axial CT technology primarily as a means to correlate CT radiation exposure for different CT scanners. Over the years, the CTDI has also come to be accepted as an index of dose. Currently, the CTDI volume (CTDI_{vol}) and dose length product (DLP) calculations are displayed on CT consoles as a means to assess scan parameter impact to deliverable dose and can be archived as a “Dose Report” with the CT examination. Though accepted as a dose descriptor, the CTDI does not measure organ dose; it is only defined for two diameters of polymethyl methacrylate (PMMA) phantoms representing the head and body and any extrapolation to effective or organ dose requires conversion factors such as those generated by Monte Carlo.²

A more direct means of estimating organ dose for patients undergoing CT examinations utilizes an anthropomorphic phantom in conjunction with thermoluminescent dosimeters (TLDs) or metal-oxide-semiconductor field-effect transistor (MOSFET);^{4,5} this is called a point dose method (PDM). Anthropomorphic phantoms are humanlike in size and weight and are designed with complex heterogeneous tissue types to emulate real patient attenuation and scatter properties. TLD and MOSFET dosimeters can be used to measure point doses in each organ throughout the phantom. However, TLD measurements are labor intensive and time consuming to use and MOSFET detectors have limited life spans. Furthermore, PDM measurements assume a homogeneous dose distribution within an organ volume when quoting an average response per organ. However, this assumption is intuitively incorrect especially for CT exposures to geometrically complex phantoms with tissues of heterogeneous attenuation coefficients. Assuming a homogeneous dose distribution could lead to an underestimation or overestimation of radiation risk in effective dose calculations for large organs such as the liver or lung that may extend out of the primary coverage area of a CT scan.

In this work, we propose a novel two-dimensional (2-D) radiochromic film dosimetry technique using GafChromic™

XRQA film. XRQA film measures high-resolution, slice-by-slice dose distribution within an organ volume. With simultaneous spatial and point dose measurements, XRQA film offers potential 2-D slice and three-dimensional volumetric CT dosimetry. Dose distribution CT dosimetry may provide a more robust means for determining radiation risk to an organ volume from CT examinations.

The following study was comprised of two main investigations: (1) Measurements were compared between XRQA film and TLDs within a geometrically simple phantom of uniform scatter and attenuation material. The simple phantom was used to verify film response accuracy and sensitivity to CT exposure geometry, i.e., with a geometry that orients the more sensitive, large surface area of the film, parallel to the main axis of the CT beam. This phantom was created by combining two CTDI head phantoms into one with the dosimeters situated in between the overall phantom halves. (2) Measurements between XRQA film and TLDs were compared within a 5 yr old pediatric anthropomorphic phantom as a means for organ dosimetry in the lungs, liver, and kidneys. For data analysis within the phantom volumes of aims (1) and (2), special software was written to individually calibrate TLD and film dosimeter responses for different tissue attenuation types; to spatially register dosimeters with respect to phantom and CT coordinate systems; and to extract dose responses from film for TLD comparison. As a comparison, TLDs were considered the gold standard and were used to measure dose in the same geometry as the XRQA films.

II. MATERIALS AND METHODS

Two CTDI head phantoms were combined, end-to-end, to form one geometrically simple phantom that provided a uniform scatter environment [Fig. 1(a)]. The CTDI head phantoms, together, measured 16 cm in diameter by 30 cm in length. Each phantom had nine probe holes that ran the length of the phantom. The unused holes were plugged using acrylic rods to remove the effects of radiation scatter in air. The phantoms were aligned to create a grid pattern shown in Figs. 2(a) and 2(b). The two most lateral holes each had an acrylic rod inserted between the two CTDI phantoms to act as an anchor to prevent either phantom from rotating with respect to the other. Once the phantoms were placed together, they were held together by tape.

A 5 yr old anthropomorphic phantom (CIRS, Norfolk, VA) was used in this study to emulate pediatric patient geometry. The phantom included 26 numbered transverse sections, each 25 mm thick [Figs. 3(a) and 3(b)]. Each slice had 5 mm diameter holes placed in predefined grid patterns [Fig. 3(c)]. The holes were numbered on the phantom [not shown

FIG. 1. (a) XRQA films were situated between two CTDI head phantoms. The two CTDI phantoms provided a uniform scattering environment for film detection. Each film was exposed to identical scan parameters, but with varying tube currents (mA). The film was used to measure the overall dose distribution uniformity from an axial CT with varied tube currents: (a) 100, (b) 200, and (c) 300 mA; note the dose color scale is different for each tube current plot. Arrows in plot (c) represent areas of higher dose due to air gaps in the CTDI phantom setup. Vertical line profiles were taken along each of the films exposed to the three different mA settings and showed an increase in dose levels from the anterior-to-posterior sides of the film. Dose levels over select areas of the films are compared to the calculated CTDI_{vol} of the given scan parameters, i.e., (f) (100–300) mA.

in Fig. 3(c)] for convenience when correlating dosimeter position with organ selection. Each organ had a number of holes within its boundaries for TLD placement. Unused holes were filled with material equivalent plugs. The phantom consisted of five tissue types: Soft tissue, lung, brain, spinal cord, and bone. The tissue types were engineered to produce photon attenuation values within 1% for bone and soft tissue substitute and 3% for lung tissue substitute over the range of 30–20 000 keV (as quoted by CIRS); hence, the phantoms anthropomorphic likeness in reconstructed CT images [Fig. 3(d)].

XRQA films (International Specialty Products, Wayne, NJ) used in this study were ~ 0.3 mm thick, had an average weighted density of 1.4 g cm^{-3} , and an atomic composition⁶ with a calculated⁷ Z_{eff} of 25.0. XRQA construction included several high- Z elements and an opaque polyester protective surface. The higher Z elements increased XRQA film's sensitivity to lower x-ray energies due to the increased photoelectric cross section. The inclusion of a white polyester film layer necessitated film digitization with a reflective densitometer.^{8,9} The XRQA film underwent solid-state polymerization in reaction to incident radiation.¹⁰ The polymerization reaction created a polymer dye complex that appeared darker in color to the eye compared to unexposed portions of the film; the reaction occurred in real-time. After

the radiation source was shut off the polymerization reaction required time to polymerize before film digitization could be made. In the case of these experiments, the films were allowed 24 h for the polymerization reaction to induce negligibly small increases in optical density (OD).

XRQA films were digitized using reflection densitometry on a flatbed scanner (Expression 10000XL, Epson America, Long Beach, CA). The radiochromic film scanner employed a xenon gas fluorescent source light and a linear charged coupled device array for film readout.¹¹ The scanner's source light illuminated only during previewing and scanning; so to "warm-up" the scanner, ten blank scans were performed in succession and then discarded. Warming up the scanner provided a more stable light source and more consistent OD readings.¹² The films were placed on the scanner in an area of high source-light uniformity, which corresponded to the center of the scanner. Each exposed film was digitized once at a resolution of 72 pixels per inch (0.4 mm pixels), with no color corrections, and image files were saved in uncompressed format. All films were digitized using 48-bit color depth with RGB file format (16-bit per color channel). The red (R of the RGB) channel was extracted for analysis in MATLAB (R2007a, Mathworks Inc., Natick, MA) since radiochromic film readout sensitivity has been shown to be optimized using a red light due to the films reported absorption

FIG. 2. (a) TLD and (b) XRQA measured dose levels are compared. TLD and film detectors were exposed with 300 mA tube current. (c) Seven locations over the area of the CTDI phantom were used to compare dosimeter responses. Initial film response levels were 45% lower than the TLD responses. The difference in film and TLD responses was consistent among high and low dose regions and therefore was correctable. The initial film response was corrected using the average percent difference calculated between the film and TLD values.

spectrum maximum at 635 nm.¹³ The final image was smoothed using a 2-D Wiener filter with a 3×3 pixel kernel.¹⁴

The films were digitized prior to being exposed and then carefully realigned on the scanner for postexposure digitization (alignment error measured to be ~ 1 mm). Care was taken to ensure that the films were always scanned in landscape mode. Once irradiated, the films darkened such that their OD exponentially increased with exposure.¹⁵ A net OD

(netOD) (Ref. 16) was calculated as the log transformation of the ratio of preirradiation (I_{pre}) and postirradiation (I_{post}) scanner light intensity values

$$\begin{aligned} \text{netOD} &= \text{OD}_{\text{post}} - \text{OD}_{\text{pre}} = \log_{10}\left(\frac{I_0}{I_{\text{post}}}\right) - \log_{10}\left(\frac{I_0}{I_{\text{pre}}}\right) \\ &= \log_{10}\left(\frac{I_{\text{pre}}}{I_{\text{post}}}\right), \end{aligned} \quad (1)$$

where I_0 was the light intensity of the flat field scan.

For film verification, $(3 \times 3 \times 1)$ mm³ Harshaw TLD-100 chips (LiF:MgTi, Thermo Eberline, LLC, Franklin, MA) were used. The TLDs were first tested for batch uniformity; all chips that deviated >2 SD from the batch mean were discarded. TLDs were annealed using a TLD-annealing furnace (168–300, Radiation Products Design, Inc., Albertville, MN) and read using a TLD reader (Harshaw 5500, Thermo Fisher Scientific, Inc., Waltham, MA) with WINREMS software. Nitrogen gas was introduced during the readout cycle to reduce nonradiation-induced signals.¹⁷

A 0.18 cc ion chamber (10X5–0.18, Radcal, Monrovia, CA), with an ion chamber monitor (9015, Radcal), was used for TLD and film calibration purposes. The ion chamber's active volume had a diameter of 14 mm and a length of 19 mm. The overall length of the chamber was 45 mm (active volume plus stem). The ion chamber anode and active volume wall material was made from graphite and was near air equivalent at the effective CT energies used in this study.

II.A. Dosimeter dose calibration

To correlate the *in-situ* TLD and XRQA film response to dose, dose calibration curves were developed. Two TLDs, one XRQA film, and an ion chamber were placed on a Styrofoam block that was extended from the CT couch so that the dosimeters were placed at the CT scanners isocenter in air [Fig. 4(a)]. Four different sets of dosimeters were exposed at varying tube current settings, 150–350 mA, at 120 kVp to provide a full range of calculated CT doses. An additional set of TLDs and film were left unirradiated for background correction. The calibration of the dosimeters occurred concurrently with the phantom study. To convert the exposure $[X(R)]$ measurements from the ion chamber to dose, the American Association of Physicists in Medicine (AAPM) TG-61 formalism¹⁸ was adopted, as was implemented by Tomic *et al.*¹⁹ The following equation was used to calculate absorbed dose to a specific medium (air) or tissue type (lung and soft tissue):

$$\text{Dose(mGy)} = 8.69 \left(\frac{\text{mGy}}{R} \right) \cdot X(R) \cdot N_x \cdot \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{tissue}}^w, \quad (2)$$

where 8.69 mGy R⁻¹ was derived from the definition of the Roentgen (R) and $\bar{W}_{\text{air}}/e$,²⁰ $X(R)$ was the ionization in air reading that had been temperature and pressure corrected, N_x was the ion chamber correction factor obtained from the University of Wisconsin Accredited Dosimetry Calibration Laboratory, and $(\mu_{\text{en}}/\rho)_{\text{tissue}}^w$ was the mean mass-energy attenuation coefficient ratio derived from tabular data provided

FIG. 3. (a) A CIRS 5 yr old anthropomorphic phantom with radio opaque markers is shown. Radio opaque markers indicate slice location of film and TLD dosimeters for dosimeter registration purposes. (b) XRQA film is shown between phantom slices. (c) An axial view of the phantom demonstrates tissue types as so indicated. The black dots represent predefined dosimeter holes where TLDs were inserted for *in-situ* measurement. (d) A corresponding axial view of the phantom in (c) is shown (soft tissue window) after CT reconstruction.

in AAPMs TG-61.¹⁸ The mean mass-energy attenuation coefficient ratios were a function of beam quality, as specified by a half value layer (HVL) measurement. The beam quality for the CT system used for dose calculations had a measured HVL of 6.8 mm Al.

Separate calibration curves were developed for lung and soft tissue types and for air (used in the case of the CTDI phantom exposure experiments) to calibrate the TLDs and film. The TLD detector response plots were fit using a linear-regression curve [Fig. 4(b)]. The average calculated coefficient of determination (R^2) for the TLD fitting functions was 0.9880. The XRQA film response plots were fit using a modified exponential function initially obtained from Ramado *et al.*¹⁵

$$\text{Dose}(\text{netOD}; a, b) = a \cdot (e^{b \cdot \text{netOD}} - 1). \quad (3)$$

A Levenberg–Marquardt algorithm was written to optimize the coefficients a and b shown in Eq. (3). The fitted functions for XRQA film are shown in Fig. 4(c) with a plot of the residual fitting error [Fig. 4(d)]; the average calculated R^2 for the radiochromic film fitting function was 0.9999. The film and TLD fitting functions were created and applied to the experimentally measured data using software written in house using MATLAB. The fitting function portion of the software code was benchmarked versus PRISM software (Version 4, Graphpad Software, La Jolla, CA) as a means to validate its accuracy.

II.B. Phantom setup and scan parameters

The CTDI and anthropomorphic phantom experimental setups consisted of three phases: (1) The TLD exposure, (2) film exposure, and (3) a phantom only exposure. The phantoms loaded with TLD and film dosimeters were scanned separately so not to introduce possible spurious results from

interdosimeter scatter. A phantom only scan was required for the analysis software to be used for organ contouring and segmentation as part of the film calibration process.

A 64 slice CT scanner (VCT, GE healthcare, Milwaukee, WI) was used for imaging the phantoms. The CT scan and reconstruction parameters used for all three scans consisted of: 120 kVp tube potential, pitch of 1 (axial scan), 40 mm detector width collimation, 5 mm reconstructed slice thickness, 1 s x-ray tube rotation time, and the small bowtie filter (head filter). Generally, most films were exposed to an x-ray tube current of 300 mA except for the CTDI phantom exposure where additional scans were performed at x-ray tube currents of 200 and 100 mA.

II.C. CTDI phantom exposure experiment

XRQA film was cut to match the shape of the CTDI phantom profile and placed between the center of the two CTDI phantoms. Once the phantom halves were in place, the probe holes were filled with acrylic rods. The rods were placed in contact with the film from both ends of the phantom. Five separate films were exposed to three different x-ray tube currents: Three CT scan acquisitions at 300 mA and one CT scan acquisition each at 200 and 100 mA; each film was exposed separately. Seven sets of three TLDs were bagged and taped into place such that the TLDs were suspended within the seven probe holes [Fig. 2(a)]. Acrylic rods were then carefully placed in contact with the TLDs from both ends of the phantom. The TLDs were exposed to one CT scan acquisition with a tube current of 300 mA.

II.D. Anthropomorphic phantom exposure experiment

Three organs were identified for dosimetry: lungs, liver, and kidneys. The lung region spanned six phantom slabs (numbers 10–15) and the kidneys and liver spanned three

FIG. 4. (a) XRQA and TLD dosimeters were placed on a Styrofoam block and placed next to an ion chamber at the CT isocenter for an in-air response-to-dose calibration. (b) The TLD dosimeter response was plotted versus dose and fit using a linear-regression curve. (c) The XRQA film was fit with a modified exponential function using a Levenberg–Marquardt algorithm for coefficient optimization and (d) the residual fitting error of the exponential fitting function was plotted.

and four phantom slabs, respectively (numbers 15–18). The lung, liver, and kidney regions were CT scanned in one contiguous interval; the interval began at the neck and ended in the pelvis region. Select locations within the anthropomorphic phantom organs were filled with two TLDs each and the remaining holes were filled with tissue equivalent material. The TLDs in each hole were set to be flush with the top of the phantom slab, so when the phantom was assembled, no pressure was placed on the TLDs. The phantom slabs containing dosimeters were marked using radio opaque fiducial markers on the anterior surface of the phantom [Fig. 3(a)]. The fiducial markers were used for TLD and film spatial registration. The anthropomorphic phantom, with TLDs *in-situ*, was carefully aligned within the CT bore using lasers and marked for subsequent repositioning for film and phantom only CT exams. After the TLD scan, the TLDs were removed and the holes that held TLDs were filled with tissue equivalent plugs. The XRQA film, which was previously cut to shape to match the different anthropomorphic phantom slabs, was placed between the phantom slabs [Fig. 3(b)]. The phantom was assembled with nine films between phantom slabs and repositioned on the CT couch. The positional dis-

placement, within the anthropomorphic phantom, between TLDs and film, was ~ 1 mm. Two sets of films, each separately exposed at an x-ray tube current of 300 mA, were measured as a means to assess experimental precision. After the CT scan of the anthropomorphic phantom loaded with film, the films were removed and the phantom was reassembled, repositioned, and CT scanned without any dosimeters *in-situ*. The phantom only CT acquisition was later used in the postprocessing procedure to extract contours of the lungs and register the TLD and film data together.

II.E. XRQA and TLD data reduction

No software existed that could perform the task of calibrating 2-D films within a heterogeneous tissue environment and spatially register and analyze multiple dosimeters at a time. Therefore, software was developed in MATLAB to perform these tasks. The following data were acquired and loaded into the software: Digitized pre-exposed and postexposed films, TLD raw data, and the reconstructed DICOM images of the phantom only scans (both anthropomorphic and CTDI). All scanned films and TLD readout results were

categorized by reconstructed slice location. Pre-exposed and postexposed films were registered and converted from pixel intensity values to netOD using Eq. (1). DICOM images from the anthropomorphic phantom scan acquisition were used to differentiate and contour lung boundaries. To differentiate and contour the soft tissue organs, mathematical coordinates of the organ locations within the anthropomorphic phantom were used to develop geometrical masks [Fig. 6(a)]. This was necessary because the anthropomorphic phantom emulated all soft tissue organs as one material and therefore there was no contrast between the liver, kidneys, and surrounding organs in the abdomen. The mathematically generated masks were registered with the reconstructed images from the phantom only CT acquisitions to indicate the location of the organs within the coordinate system of the DICOM images. The phantom only DICOM images were registered with the films and used to contour the lungs and soft tissue organ masks. Once registered, the film and DICOM images shared the same coordinate system. The information of the spatial locations of the organs was taken from the DICOM images and used to extract the dosimetric information from the films by overlaying the DICOM contours onto the films. The netOD from the extracted film data, corresponding to regions of lung and soft tissue, were then calibrated separately using Eq. (3) preserving the attenuation specific attributes of each organ. Since the CTDI phantom was made of homogeneous PMMA material, a simple external contour around the phantom was used to calibrate the film. The results from the CTDI phantom were quoted with respect to air attenuation. To compare TLD readings with XRQA film measurements, for both the CTDI and anthropomorphic phantom experiments, the films were registered with digitized images of the phantom slabs to match the location of TLDs with spatial locations in the film coordinate system. Once registered, regions of interest (ROIs) were extracted from the film at the locations that corresponded to the TLD positions and were plotted for comparison.

III. RESULTS

III.A. Film/TLD dose comparison from CTDI phantom experiment

Five films were separately exposed when placed within the CTDI phantom; one film each was CT scanned at 100 and 200 mA, [Figs. 1(b) and 1(c)], respectively, and three films were CT scanned at 300 mA and averaged [see Fig. 1(d)]. Generally, the films measured a greater dose along the anterior portion of the film than compared to the posterior portion. However, the film exposed at 200 mA exhibited a high dose region in the posterior portion of the film. These regions of high dose are located in the surrounding area of grid numbers 5 and 6, indicated by the arrows in Fig. 1(c), and were due to the acrylic rods not being in contact with the film, thus leaving an air gap between the film and the rods (which was evident from a review of the DICOM images, not shown). The air pocket created a low density region that allowed more radiation to strike the film in the regions surrounding grid numbers 5 and 6 and thus increase the dose.

The difference in anterior and posterior dose levels was verified using line profiles taken from the anterior-to-posterior portions of the films, as taken along a representative dashed line in Fig. 1(c). The line profiles measured a difference in anterior from posterior dose up to 16% for films exposed at 200 and 300 mA and 13% for the film exposed at 100 mA [Fig. 1(e)]. An interesting point is that the $CTDI_{vol}$ values for 100–300 mA film exposures closely correlated with the film values [Fig. 1(f)]; its correlation was calculated to be within 10%. A slightly better agreement between the three exposed films and the $CTDI_{vol}$ values was measured when only considering the more uniform region of the film in the anterior portion of the film corresponding to grid hole numbers 1–3; its correlation was calculated to be within 4%.

As a measure of film accuracy, the film and TLD values were compared [Fig. 2(c)]. The average difference in magnitude of dose responses was calculated to be 45% ($SD \pm 2\%$). The difference in film and TLD dose response was due to the lower response sensitivity of the film that arises when the film is oriented with its large surface area parallel to the main axis of the CT beam, which findings are in good agreement with a recent publication by Rampado *et al.*²¹ The consistency of the response difference at both low and high doses allowed for a tissue specific correction to be applied. Once corrected, the average film response agreed to better than 3% ($SD \pm 2\%$) of similarly located TLDs [Fig. 2(c)].

III.B. Organ dose measurement using 5 yr old pediatric anthropomorphic

The XRQA films were registered to the reconstructed CT images of the anthropomorphic phantom [Figs. 5(a) and 6(a)] and the area of the film corresponding to the organs of interest in the CT images was extracted and converted to dose using the film dose fitting functions plotted in Fig. 4(c). The film measured lung dose response is plotted in Fig. 5(b) and the dose responses corresponding to the liver and kidneys are plotted in Fig. 6(b). The dose levels for both the thorax and abdomen regions are displayed in color wash and represented by the same dose scale measured in mGy. The magnitude of the dose reported here was not meant to be clinically realistic for pediatric scan protocols. The tube current (mA) parameter was manually set high to provide large beam fluence on the film to maximize the precision of the film results.

In reconstructed slices corresponding to phantom slabs 10–14, pairs of TLDs were placed within the lung region to verify film lung response. The location and identification of the 36 TLDs are shown in Fig. 5(b). For slice 15, the phantom slab did not have any holes within the lung region for TLD placement and, therefore, the film lung response was not verified. Similarly, in slices 15–18, 25 TLDs were placed within regions of the liver and 28 TLDs were placed within the kidneys [Fig. 6(b)]. The film and TLD dose responses for all three organs are plotted in Fig. 7. The error bars on the TLD and film measurements represent the deviation (one SD) in response of two TLDs per phantom hole and film measurements during two separate but identical experiments

FIG. 5. XRQA film was placed in between six slices in the thorax region of a 5 yr old anthropomorphic phantom and CT scanned. The reconstructed CT slices of the phantom are shown in (a). The lung regions within the reconstructed CT images were contoured and spatially correlated with the XRQA film lung regions by film/phantom registration. The lung regions of the exposed XRQA films were extracted and are shown in (b). Film dose responses (mGy) are shown in color wash. Separately, TLDs were placed within the lung volume, as indicated in (b), and the phantom and TLDs were CT scanned.

held sequentially. All film measurements correspond in space to the location of the TLD positions. A difference in film response and TLDs indicated a similar consistent difference in response for the lung, 28% ($SD \pm 8\%$), and liver and kidneys, 15% ($SD \pm 4\%$), as was seen by the CTDI phantom experiment; thus allowing for a tissue specific correction to be applied. Once corrected, the average film response in the lung tissue agreed to better than 3% ($SD \pm 3\%$), for the liver 5% ($SD \pm 3\%$), and for kidneys 4% ($SD \pm 3\%$) (Fig. 7).

IV. DISCUSSION

The lower film response, as indicated previously, was generally consistent in the low and high dose regions, as is seen by the similar trends in film and TLD responses (Fig. 7). This allowed for tissue specific correction factors. The correction factors were calculated for each tissue type by measuring the average difference in response from all of the TLD and corresponding film values throughout the entire

FIG. 6. XRQA film was placed in between three slices in the abdomen region of a 5 yr old anthropomorphic phantom and CT scanned; slice 15 is a carryover from the thorax region but had liver involvement and is therefore shown here. The reconstructed CT slices of the phantom are shown in (a). Unlike the lung regions that were easily contoured, the liver and kidneys were not discernable in the reconstructed images because they were both created to emulate the soft tissue as were the other vital organs in the abdomen region. Because there was no contrast in the phantom for direct contouring, mathematical masks were created from digitized images of the actual phantom and overlaid with the DICOM images. The liver was represented in magenta and the kidneys in green. The films were spatially registered with the DICOM images and the areas within the film corresponding to the liver and kidneys were extracted and are shown in (b). Film dose responses (mGy) are shown in color wash and are on the same scale as lung dose responses shown in Fig. 5. TLDs placed within the liver and kidney volumes are indicated in (b).

organ volume. The average difference of the two responses was used to globally correct the initial film dose values for all films located within the volume of the specific organ. This volumetric approach to correcting the film response was more consistent for the two soft tissue organs, the liver and kidney. A possible reason was that within the abdomen region, the scatter conditions were more consistent from slice-to-slice because of the consistency in body habitus shape and homogeneity of the emulated abdomen tissue [Fig. 6(a)]. Within the thorax region, the lung dose varied more, slice-to-slice, due to the change in scatter conditions from the varying difference in the ratio of lung and soft tissue area. The difference in scatter conditions from one slice of the lung to the next possibly caused the volume correction factor to under correct the film lung response in slice 12 (TLDs 23–30) in Fig. 7. Considering this point, a slice-by-slice correction method may be more appropriate. This is accom-

FIG. 7. TLD and XRQA film dose responses are compared for the (a) lung, (b) liver, and (c) kidneys. The XRQA film dose values were extracted using ROIs centered over the corresponding locations of the TLDs in the phantom [Figs. 5(b) and 6(b)]. Film and TLD values maintain a similar response pattern due to identical scan parameters and scatter geometries. Film values corrected using the tissue specific correction factors generally maintained similar dose response trends since film and TLD readings were consistently offset in both the high and low dose regions. Error bars for film and TLD are a measure of deviation for the two TLDs and two films measured per location (quoted at one SD).

plished by averaging the difference of the film and TLDs per slice and using the percent difference in that slice only to correct the film.

When considering the impact of radiation measurement techniques on risk assessment for organ based dosimetry, we propose that there is a clear need for 2-D distributive dose information. With the introduction of radiochromic film as a CT radiation dosimeter, CT dosimetry takes on another dimension of vital information. Instead of a simple point dose measurement for the organ, CT film dosimetry now relates the amount and 2-D distribution of the dose in the organ volume. With dose distribution measurements, nonuniform dose deposition within an organ volume can be used to better provide a more complete dosimetric analysis for organ risk and CT scan protocol development as a means to mitigate increasing trends of CT dose. As a demonstration of the importance of radiochromic film dosimetry, the dose to the lung [Fig. 5(b)] is greater along the anterior medial portions of the bilateral lungs, which measurements are not captured by the placement of any TLDs. The increase in anterior dose measured by the film in the anthropomorphic phantom is supported by the CTDI phantom exposure results. Unfortunately, no predefined hole in the anthropomorphic phantom

grid configuration would have allowed for the TLDs to capture this spatial distributed dose information; see Fig. 3(c) as one example of the phantom dosimeter grid spacing within the lung. Thus, using a PDM technique will be intrinsically limited when measuring inhomogeneous dose distributions. This PDM limitation is also evident by the fact that the TLD measurements are shown to underestimate the maximum dose to the lung volume by 15%. TLD measured maximum dose was (70.4 ± 1.1) mGy (TLDs 42 and 43 located in slice 14) and the measured minimum dose was (52.8 ± 1.3) mGy (TLDs 11 and 12 located in slice 10), whereas the film measured maximum dose was (82.5 ± 0.3) mGy (slice 14, anterior patient right lobe) and minimum dose was (52.7 ± 0.3) mGy (slice 10, inferior middle patient right lobe). The TLD results were not inaccurate; they were just not located properly to measure the maximum dose in the lung volume.

The measured dose inhomogeneity to the organs is a function of the geometric complexity of the body and the different attenuating and scattering tissue types within that geometry. An argument can be made from the evidence on hand that to use a point dose dosimeter to measure dose within a

volume and assume that dose is indicative of the entire volume is incorrect. Therefore, it would be incorrect to alter CT protocols and develop scan protocols based on a limited measurement methodology.

Future considerations for this work may focus on reviewing the consistency of the tissue specific correction factors for different detector collimation configurations, i.e., varying the scan parameters to verify how the subsequent effects in changing scatter conditions might affect the use of one single correction factor per tissue type. Also, the eventual goal of this work will be to introduce helical scan parameters for more clinically relevant scatter geometries and dose distribution analysis. Finally, to increase the dose resolution in the z-axis (body length direction) measurements on a modified anthropomorphic phantom with slice thicknesses less than 25 mm thick will be required.

V. CONCLUSION

In this work, we proposed a novel radiochromic film dosimetry technique, using GafChromic™ XRQA film, for organ dosimetry within a pediatric anthropomorphic phantom. XRQA film was used to measure a spatially high-resolution slice-by-slice dose distribution within an organ volume. First, the film accuracy and sensitivity to a CT scan geometry was validated using TLDs within a modified CTDI phantom. The film measurements consistently under responded in the high and low dose regions when compared to TLD results. This allowed for the film to have a correction factor applied. Once applied, the film agreed to better than 3% ($SD \pm 2\%$) for the CTDI phantom. The under response of the film was attributed to the lower detection sensitivity of an x-ray beam when the film is oriented such that its large surface area is parallel to the x-ray beam's main axis. Measurements between XRQA film and TLDs were then compared within a 5 yr old pediatric anthropomorphic phantom in a similar methodology as the CTDI phantom experiment. To analyze the data within the complex heterogeneous anthropomorphic phantom, software was developed to individually calibrate TLD and film dosimeter responses for different tissue attenuation factors; to spatially register multiple dosimeters with respect to phantom and CT coordinate systems; and to extract dose responses from film for TLD comparison. The subsequent results of the dose measured for the lungs, liver, and kidneys also indicated a consistent geometrical sensitivity of the film detection within the anthropomorphic phantom. Once corrected, the film response agreed to better than 3% ($SD \pm 3\%$) for lungs, 5% ($SD \pm 3\%$) for liver, and 4% ($SD \pm 3\%$) for kidneys. Additionally, high-resolution dose distributions were measured within the anthropomorphic phantom organ volumes. The film dosimetry measured a nonuniform deposition of dose to lung, liver, and kidney volumes, the extent of which the TLD detectors were unable to measure. In fact, the film measured a 15% greater maximum dose in the lung tissue than the TLDs. A 2-D film measurement was essential in this instance to accurately locate the maximum dose and measure the extent of the dose distribution within the organ volumes. XRQA radiochromic film has

been shown to provide a novel and accurate organ dosimetry within a CT scan volume. Dose distributive CT dosimetry may provide a more robust means for determining radiation impact from a CT scan to an organ volume.

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