Quantitative megavoltage radiation therapy dosimetry using the storage phosphor KCl:Eu^{2+}

Zhaohui Han, Joseph P. Driewer, Yuanshui Zheng, Daniel A. Low, and H. Harold Li^{a)}

Department of Radiation Oncology, Washington University School of Medicine, 4921 Parkview Place, Campus Box 8224, St. Louis, Missouri 63110

(Received 23 March 2009; revised 15 June 2009; accepted for publication 16 June 2009; published 15 July 2009)

This work, for the first time, reports the use of europium doped potassium chloride (KCl:Eu^{2+}) storage phosphor for quantitative megavoltage radiation therapy dosimetry. In principle, KCl:Eu^{2+} functions using the same photostimulated luminescence (PSL) mechanism as commercially available $\text{BaFBr}_{0.85}\text{I}_{0.15}\text{:Eu}^{2+}$ material that is used for computed radiography (CR) but features a significantly smaller effective atomic number—18 versus 49—making it a potentially useful material for nearly tissue-equivalent radiation dosimetry. Cylindrical KCl:Eu^{2+} dosimeters, 7 mm in diameter and 1 mm thick, were fabricated in-house. Dosimetric properties, including radiation hardness, response linearity, signal fading, dose rate sensitivity, and energy dependence, were studied with a laboratory optical reader after irradiation by a linear accelerator. The overall experimental uncertainty was estimated to be within $\pm 2.5\%$. The findings were (1) KCl:Eu^{2+} showed satisfactory radiation hardness. There was no significant change in the stimulation spectra after irradiation up to 200 Gy when compared to a fresh dosimeter, indicating that this material could be reused at least 100 times if 2 Gy per use was assumed, e.g., for patient-specific IMRT QA. (2) KCl:Eu^{2+} exhibited supralinear response to dose after irradiation from 0 to 800 cGy. (3) After x ray irradiation, the PSL signal faded with time and eventually reached a fading rate of about 0.1% / h after 12 h. (4) The sensitivity of the dosimeter was independent of the dose rate ranging from 15 to 1000 cGy/min. (5) The sensitivity showed no beam energy dependence for either open x ray or megavoltage electron fields. (6) Over-response to low-energy scattered photons was comparable to radiographic film, e.g., Kodak EDR2 film. By sandwiching dosimeters between low-energy photon filters (0.3 mm thick lead foils) during irradiation, the over-response was reduced. The authors have demonstrated that KCl:Eu^{2+} dosimeters have many desirable dosimetric characteristics that make the material conducive to radiation therapy dosimetry. In the future, a large-area KCl:Eu^{2+} -based CR plate with a thickness of the order of a few microns, created using modern thin film techniques, could provide a reusable, quantitative, high-resolution two-dimensional dosimeter with minimal energy dependence. © 2009 American Association of Physicists in Medicine. [DOI: 10.1118/1.3171687]

Key words: storage phosphor, dosimeter, IMRT, radiation therapy dosimetry

I. INTRODUCTION

Modern radiation therapy has advanced considerably in the past decade with the development of improved imaging and conformal dose delivery techniques [e.g., intensity modulated radiation therapy (IMRT)].^{1,2} With the subsequent increased complexity of treatment planning and dose delivery, there has been a growing demand for multidimensional quantitative dose delivery validation.^{3,4} Radiographic film has become the *de facto* standard for two-dimensional (2D) dose distribution measurements, with three-dimensional dose distribution validation conducted using stacked film measurements.^{3–16} As diagnostic radiology and radiation oncology clinics move toward electronic imaging, the reduced availability of radiographic film and film processors is becoming increasingly problematic. Furthermore, since film is fundamentally a single use detector, it cannot be reliably calibrated. Quantitative dosimetry with film requires the acquisition of a sensitometric curve each time a dosimetric measurement is made, but this practice is based on the assumptions that each individual film has the same response as

others in the batch and that the processor performance does not change from film to film. In this regard, near tissue-equivalent radiochromic films^{17–20} suffer from the same disadvantage in that they are not reusable. Therefore, an alternative, quantitative, reusable, high-resolution planar dosimeter is needed for IMRT commissioning and quality assurance.

In 2005, Olch evaluated commercial computed radiography (CR) plates made of $\text{BaFBr}_{0.85}\text{I}_{0.15}\text{:Eu}^{2+}$ for digital 2D megavoltage beam dosimetry.²¹ One advantage of CR plates is that they allow the physicist to monitor the long-term properties and response of the plates as dosimeters. Olch concluded that, with proper procedures, the $\text{BaFBr}_{0.85}\text{I}_{0.15}\text{:Eu}^{2+}$ -based storage phosphor plate (SPP) could be used for quantitative megavoltage beam dosimetry, with the added benefit that the readout system could be used for both imaging and dosimetry. While the commercial SPP system has many desirable dosimetric properties, current SPP systems are based on $\text{BaFBr}_{0.85}\text{I}_{0.15}\text{:Eu}^{2+}$ material, which has high-Z components (effective $Z=49$). This leads

FIG. 1. (a) In-house fabricated KCl:Eu²⁺ dosimeters. Dosimeters are 7 mm in diameter and 1 mm thick. Also shown is a holder which attaches a dosimeter to the integrating sphere during readout. (b) Readout system. Photostimulated luminescence signals are collected by an integrating sphere and detected by a photomultiplier tube. The stimulation source is either a yellow He–Ne laser or a quartz tungsten–halogen lamp coupled with a monochromator for stimulation wavelength selection.

to strong photon energy dependence for large field sizes and depths. A reusable dosimeter with a more waterlike response is desirable.

In 1993, Nanto *et al.*²² observed photostimulated luminescence (PSL) emission in europium doped potassium chloride (KCl:Eu²⁺) that was exposed by a diagnostic x ray beam. The magnitude of the signal, however, was approximately one-tenth that of BaFBr_{0.85}I_{0.15}:Eu²⁺. The decrease in sensitivity was due to the lower effective *Z* of KCl:Eu²⁺ compared to that of BaFBr_{0.85}I_{0.15}:Eu²⁺ (18 versus 49, respectively). We hypothesize that since greater dose is delivered by radiation therapy beams, KCl:Eu²⁺ will still produce an excellent signal for the purposes of dosimetry. Additionally, since the effective *Z* is closer to that of water, the KCl:Eu²⁺ storage phosphor should have a more waterlike response to beams commonly employed in radiation oncology.

The present work, for the first time, reports the use of KCl:Eu²⁺ photostimulable storage phosphor for quantitative radiation therapy dosimetry, thereby laying the groundwork for future development of a novel planar dosimeter. The prototype dosimeter used in this study is a point dosimeter with a physical makeup similar to thermoluminescent dosimeter (TLD) chips. We demonstrate that this novel dosimeter has many desirable dosimetric characteristics that make the material conducive to radiation therapy dosimetry. In the future, a large-area 2D KCl:Eu²⁺ plate created using modern thin film techniques could provide a quantitative high-resolution dosimeter with nearly water-equivalent response that can be used in existing geometric phantoms and that can also be reused so that dosimetric properties can be determined and monitored by the physicist.

II. MATERIALS AND METHODS

II.A. Dosimeters and phantom

Cylindrical KCl:Eu²⁺ dosimeters, 7 mm in diameter and 1 mm thick, shown in Fig. 1(a), were fabricated in-house. The phantom consisted of solid water slabs (SW-457, Gammex RMI, Middleton, WI) with areas of 40 × 40 cm², stacked to yield 30 cm of thickness. A 5 mm thick slab with

a linear array of holes machined 7.5 mm in diameter and 2 mm in depth served to host the dosimeters during irradiation. Central axis measurements were acquired in this study, thus the dosimeter was placed at the center hole. Prior to irradiation, the dosimeters were optically bleached (annealed) for 5 s using a 500 W tungsten-halogen lamp.

II.B. KCl:Eu²⁺ dosimeter readout

Except for PSL stimulation and emission spectra measurements, the dosimeters were read using the setup illustrated in Fig. 1(b). The stimulation power was supplied by a 100 W quartz tungsten-halogen lamp (Newport, Stratford, CT). The stimulation wavelength of 570 nm was selected using a motorized monochromator (Cornerstone 130, Newport) with a grating of 1200 lines/mm. The monochromator had an integrated shutter that was controlled through a General Purpose Interface Bus (GPIB) interface and used to switch the stimulation light on and off. The stimulation light was chopped with an optical chopper (model SR540, Stanford Research System, Sunnyvale, CA) set to 80 Hz in order to provide a reference signal to a lock-in amplifier for phase sensitive detection. The stimulation light was focused by a series of lenses and directed to an integrating sphere with multiple ports (LabSphere, North Sutton, NH). The KCl:Eu²⁺ dosimeter was mounted on a customized cap with an aperture of 7 mm [Fig. 1(a)] on a port opposite the photomultiplier tube (PMT) (Hamamatsu, Bridgewater, NJ) which collected and amplified the PSL signals. The entire sensitive area of the dosimeter was uniformly illuminated by the stimulation light. The gain of the PMT was controlled through a dc power supply (model PS310, Stanford Research System) that supplied high voltage up to 1.25 kV. Between the integrating sphere and the PMT tube was a set of Schott glass bandpass filters, a BG-3 and a BG-39, each 3 mm thick. The filter combination had a maximum sensitivity at 420 nm. The PMT output current was converted into voltage that was measured using a dual-channel lock-in amplifier (model SR830, Stanford Research System) and recorded on a personal computer. The system was controlled through a GPIB interface. A typical measurement involved opening the shut-

ter, waiting 1 s for the signal to stabilize, and then taking ten consecutive readings at a time interval of 100 ms. The mean of the ten readings was recorded as a measurement.

Background signals were measured by placing unirradiated dosimeters on the integrating sphere. The background signals consist of a dc component and a random noise signal. The dc signal was largely due to the leakage of the ambient light (e.g., from a computer monitor) through the glass filters, luminescent centers generated by cosmic radiation, and the bleaching lamp, as well as the PMT dark current. This dc signal can be as large as a few percent of the total signal, depending on the dose and stimulation power and can be removed by subtraction. The background noise, which is thermal and electronic in origin, ultimately defines the minimum detectable limit. In this study, for a typical dose of 100 cGy, a signal-to-noise ratio of greater than 1000 can be easily achieved.

Dosimeter readout was examined through the following tests.

II.B.1. Stimulation and emission spectra

The stimulation spectra were obtained by scanning the monochromator between 450 and 700 nm in 1 nm increment, while the PSL signals were collected through a narrow bandpass filter (FB420-10, Newport). The spectra were then corrected for the diffraction efficiency of the grating and the stimulation power as a function of wavelength, as measured by a calibrated silicon photodiode (model FDS100-CAL, Thorlabs, Newton, NJ).

In order to obtain the emission spectrum, a 5 mW, 594 nm yellow He–Ne laser (Melles Griot, Covina, CA) was used in place of the halogen lamp for stimulation. The laser power was attenuated so that the PSL signal decrease due to continuous stimulation during a typical scan time of 30 s was about 10%. The monochromator was placed between the integrating sphere and the PMT, which was fitted with a BG-3 and a BG-39 filter (3 mm each). The emission spectrum was obtained by scanning the monochromator from 300 to 540 nm with a 0.5 nm step and correcting for signal depletion and spectral response of the PMT.

II.B.2. Readout signal depletion

A KCl:Eu²⁺ dosimeter stores dose information in the form of trapped electrons and holes formed during ionizing irradiation. The readout process involves liberating the trapped electrons that subsequently migrate to recombination centers (Eu), where they recombine with holes under the emission of PSL photons.²³ Dose information can be read out repeatedly by multiple stimulations using the current setup; however, each subsequent reading will have slightly decreased signal strength due to the partial depletion of the trapped electrons by previous readouts. The readout signal loss was measured by taking 30 consecutive readings at an interval of 10 s on a single dosimeter 13 h after irradiation.

II.B.3. Signal fading

An important characteristic of a storage phosphor is the signal fading over time.²³ To measure this property, KCl:Eu²⁺ dosimeters were irradiated by a dose of 200 cGy and read repeatedly for 24 h, beginning immediately after irradiation. Each reading was corrected for the reading signal depletion as measured above.

In general, a fixed time delay between irradiation and readout was used to minimize errors associated with variable signal fading. In order to measure the impact of fading, a batch of eight dosimeters were exposed to a 6 MV beam up to 400 cGy and were read at delay times of 20, 44, and 68 h. The results were normalized to the readings at 100 cGy and compared.

II.C. Dosimetric characteristics

The x ray beams used in this study had nominal energies of 6, 10, and 18 MV. Electron beams had nominal energies of 6, 9, 12, 16, and 20 MeV. The beams were generated by a Varian 23EX (Varian Medical Systems, Palo Alto, CA) or an Elekta Precise (Elekta, Norcross, GA) linear accelerator. The dosimeter plane was oriented perpendicular to the beam central axis for all irradiations. During irradiation, the accelerator output was monitored using a 0.6 cm³ Farmer-type ionization chamber (PTW N23333, Friedberg, Germany) inserted into a 40×40×3 cm³ solid water slab placed 7 cm below the dosimeter plane. The nominal dose rate was 600 MU/min, unless otherwise noted.

Due to the limited availability of the linear accelerators, and to provide consistent experimental conditions, the irradiations were performed during evenings. The irradiated dosimeters were kept in the dark at room temperature and read on the next day. Before and after each irradiation session, the dosimeters were irradiated to 200 cGy in an open field in order to determine the relative sensitivity, or “chip factor,” of each dosimeter. These two sets of chip factors were usually consistent within 2% for each chip, and the averages were used to correct for the sensitivity variation among dosimeters.

The major sources of experimental uncertainty were the positioning error of dosimeters during readout and the stability of the stimulation lamp. Measurements were found to be sensitive to the mounting of the dosimeters on the integrating sphere. The fluctuation of the stimulation power was detected to be up to 1% during 20 h monitored by a Si photodiode. For all measurements, each dosimeter was read two or three times, including removing the dosimeter from the integrating sphere port and remounting. The readings were corrected for partial depletion, averaged, and corrected for chip factor. The standard deviation of repeated readings was taken as the experimental error.

The following dosimetric tests were conducted.

II.C.1. Radiation hardness

Three dosimeters were first irradiated with 50, 100, and 200 Gy by the 6 MV beam. These dosimeters were subsequently bleached, irradiated to 200 cGy, and read.

II.C.2. Dose response

Twelve dosimeters were irradiated individually up to 800 cGy with an SSD of 90 cm, depth of 10 cm, and field size of 20×20 cm² using 6, 10, and 18 MV beams.

II.C.3. Beam energy dependence

For megavoltage x ray beams, dosimeters were irradiated using 6, 10, and 18 MV photons at an SSD of 90 cm, a depth of 10 cm, and a field size of 20×20 cm². For the electron beams, the dosimeters were irradiated using 6, 9, 12, 16, and 20 MeV electron beams with the dosimeters placed at the respective depths of maximum dose ($d_{\max}$).

II.C.4. Dose rate dependence

The dose rate dependence was investigated using the 6 MV beam. The dosimeters were placed at $d_{\max}$ (1.5 cm) with a field size of 10×10 cm² and an SSD of 100 cm. Nominal dose rates of 100, 200, 300, 400, 500, and 600 cGy/min (based on the same number of MU/min) were delivered by changing the accelerator repetition rate. A dose rate of 15 cGy/min was delivered by irradiating the dosimeter underneath fully closed multileaf collimators (MLCs). A dose rate of 16.5 cGy/min was achieved at an SSD of 424.5 cm. A higher dose rate of 1000 cGy/min was achieved by reducing the SSD to 77 cm. At each dose rate, a dose of 100 cGy was delivered to the dosimeter.

II.C.5. Field size and depth dependence

Varying field sizes and depths altered the incident fluence spectra, i.e., the scatter-to-primary x ray ratios, at the dosimeter plane.¹³ The dosimeters were irradiated at the isocenter to 200 cGy (SAD=100 cm) at four depths (5–20 cm) for each of five square field sizes (5×5 – 25×25 cm²). A sensitometric curve for the dose range of 170–250 cGy was obtained at a depth of 10 cm and a field size of 10×10 cm². The ratio of the measured dose using this calibration curve to the dose measured by an ionization chamber was used to indicate the magnitude of dose measurement error. For comparison, the measurements were repeated using Kodak XV and Kodak EDR2 radiographic films and the Agfa MD10 (Eastman Kodak, Rochester, NY) CR plate. The films were processed using a diagnostic quality film processor (Kodak RP X-Omat Processor, Eastman Kodak Co., Rochester, NY) and the optical densities were measured using a calibrated manual densitometer (Digital Densitometer II, Victoreen). The Agfa MD10 CR plate was made of BaFBr_{0.85}I_{0.15}:Eu²⁺.²³ Small (7 mm in diameter) chips were cut from a CR plate. The chips were irradiated in the same manner as the KCl:Eu²⁺ chips and read using the in-house reader described in Sec. II B.

FIG. 2. Emission and stimulation spectra of KCl:Eu²⁺. When stimulated, the KCl:Eu²⁺ dosimeter emits intense photostimulated luminescence centered at 420 nm. The maximum stimulation efficiency occurs at 570 nm (the peak of the stimulation spectrum), which was chosen as the readout wavelength in this study.

II.C.6. Field size and depth dependence with low-energy photon filters

It has been shown that low-energy photon filters placed inside the phantom minimize the over-response of radiographic film by selectively attenuating low-energy scattered photons before they can interact with the film.^{21,24,25} The above field size and depth dependence measurements were repeated with two 0.3 mm thick lead foils sandwiching the dosimeters, separated by 6 mm solid water material upstream and 3 mm downstream.

II.C.7. Angular dependence

For angular dependence measurements, two identical half-cylinders of 2 cm in radius and 8 cm in length were machined from a solid water slab. A hole of 7.5 mm in diameter and 1.5 mm in depth was machined at the center of one of the half-cylinders to hold the dosimeters. The two half-cylinders were taped together to form a cylinder in such a way that the dosimeters were near the geometrical center of the cylinder. During irradiation, the gantry was set at 90° so that the beam axis was parallel to the floor. The cylinder was placed vertically on a Styrofoam block which was marked so that the angle of the dosimeter relative to the beam axis could be easily set. The long axis of the cylinder was the rotational axis. This setup ensured a homogeneous depth in all directions. A field size of 10×10 cm² was used. The relative sensitivity of the dosimeters was determined for angles up to 90° with an increment of 15°.

III. RESULTS

III.A. KCl:Eu²⁺ dosimeter readout

III.A.1. Stimulation and emission spectra

Figure 2 shows the emission and stimulation spectra of the KCl:Eu²⁺ dosimeter after x ray irradiation. The dosimeter emitted intense PSL centered at 420 nm when stimulated

FIG. 3. The depletion of trapped electrons due to repeated readings. A KCl:Eu²⁺ dosimeter was kept in the dark for 13 h after x ray irradiation. Thirty consecutive readings were then made with an interval of 10 s. A linear fit to the data indicated a depletion rate of 0.27% per reading, the value used to correct readings in measurements (e.g., Fig. 4). Note that a smaller depletion rate can readily be achieved by attenuating the stimulation power using gray filters.

with a yellow laser (594 nm). The maximum stimulation efficiency occurred at 570 nm, which was subsequently selected as the readout wavelength in this study. These findings were consistent with earlier reports.^{22,26}

III.A.2. Readout signal depletion

Figure 3 shows 30 consecutive readings taken at 10 s intervals 13 h after irradiation. The total read time was about 5 min. As shown later, the signal fading was negligible during this 5 min period. The signal decrease was therefore attributed solely to partial depletion of trapped electrons.

A linear fit of the 30 readings had a slope of -0.0027 , indicating that each reading depleted approximately 0.27% of the existing trapped electrons. It should be noted that the stimulation power can be straightforwardly altered by applying gray filters. Therefore, a smaller depletion rate could be readily achieved at the expense of signal strength.

III.A.3. Signal fading

Figure 4 shows the fading characteristics of the prototype KCl:Eu²⁺ dosimeter. The fading curve was corrected for the partial depletion of 0.27% per readout as determined previously. 63% of the original signal remained after 12 h, and the signal eventually reached a fading rate of about 0.1%/h. Also shown in Fig. 4 is the fading curve for a commercial Agfa MD10 CR plate. The stimulation wavelength was 620 nm, corresponding to the maximum stimulation efficiency for BaFBr_{0.85}I_{0.15}:Eu²⁺ material. Based on 30 consecutive readings, a depletion rate of 0.07% per reading was obtained, which was corrected for in the fading curve. Compared to the Agfa MD10 CR plate, KCl:Eu²⁺ showed slower fading rates.

FIG. 4. Signal fading over time. The fading of KCl:Eu²⁺ (diamonds) consists of a fast component immediately after irradiation and a slow component afterwards. The signal eventually reaches a fading rate of about 0.1%/h. 63% of the original signal remains after 12 h. For comparison, the fading of a commercial Agfa MD10 CR plate is also shown (circles). KCl:Eu²⁺ dosimeters exhibit improved fading characteristics.

III.B. Dosimetric characteristics

III.B.1. Radiation hardness

In order to serve as a reusable dosimeter, the radiation resistance of the active material has to be sufficiently high. As shown in Fig. 5, there was no significant change in the stimulation spectra after irradiation to 100 and 200 Gy, respectively, compared to a fresh KCl:Eu²⁺ dosimeter. The slight redshift could be caused by the increased creation of M(Cl⁻) centers, the aggregate center of two neighboring chlorine ion vacancies occupied by two electrons.²⁷ A 15% loss in sensitivity was observed for an accumulated dose of 50 Gy. Little further loss was detected up to 200 Gy. The sensitivity loss could be attributed to an agglomeration of Eu²⁺, leading to luminescence quenching.

FIG. 5. Radiation hardness. Three KCl:Eu²⁺ dosimeters were first given 0, 100, and 200 Gy, respectively, from a 6 MV beam. They were subsequently bleached, irradiated to 200 cGy, and read. No significant change is observed in the stimulation spectra indicating that KCl:Eu²⁺ has satisfactory radiation resistance.

FIG. 6. KCl:Eu²⁺ dosimeters' response to absorbed x ray dose. (a) Readout was made 20 h after irradiation. The dose response shows a supralinear behavior which can be fitted with a second-order polynomial. The response is independent of the beam energy within 2%. (b) Readout was made at different delay times after irradiation to a 6 MV beam. The PSL amplitude was normalized to its value at 100 cGy. The normalized response shows no dependence on delay time. The error bars represent 1 standard deviation of repeated measurements.

III.B.2. Dose response

Figure 6(a) shows the dosimeter response to x ray dose for the 6 and 18 MV beams. The two curves agreed with each other to within 2%. Similar to most TLDs, KCl:Eu²⁺ dosimeters exhibited a supralinear behavior in dose response.²⁸ The response curve was fitted to a second-order polynomial,

$$\text{Signal (V)} = 2.741 \times 10^{-4} \times \text{Dose (cGy)} + 1.062 \times 10^{-7} \times \text{Dose}^2 \text{ (cGy}^2\text{)},$$

which was plotted as the solid line in Fig. 6(a).

As mentioned above, a fixed delay time of 20 h between irradiation and reading was used. Figure 6(b) shows the measurements, normalized to the readings at 100 cGy, for a batch of eight dosimeters exposed to the 6 MV beam up to 400 cGy and read at different delay times of 20, 44, and 68 h. As the delay time increased, the PSL signal decreased following the fading curve (Fig. 4); however, the normalized curves agreed with each other within the experimental uncer-

FIG. 7. KCl:Eu²⁺ dosimeter's relative sensitivity versus beam mode and energy. The sensitivities are normalized to their respective average values. For both x ray (a) and electron (b) beams commonly available from a multimodality linear accelerator, the dosimeter's sensitivity is independent of the beam energy within $\pm 2.5\%$. The error bars represent 1 standard deviation of repeated measurements.

tainty, indicating no delay time effect. Therefore, absolute doses can be deduced from readings at any delay time provided the fading characteristics are known.

III.B.3. Beam energy dependence

The sensitivity dependence on nominal incident beam energy is summarized in Fig. 7. Within $\pm 2.5\%$ the KCl:Eu²⁺ dosimeters showed no energy dependence for either open field x rays or megavoltage electrons commonly available from a multimodality linear accelerator.

III.B.4. Dose rate dependence

Figure 8 summarizes the sensitivity variations in KCl:Eu²⁺ dosimeters as a function of dose rate after irradiation by the 6 MV beam. There was no variation in sensitivity for dose rates ranging from 15 cGy/min (underneath a fully closed MLC or at large SSD) to 1000 cGy/min (at an SSD of 77 cm) within $\pm 2.5\%$.

FIG. 8. KCl:Eu²⁺ dosimeter's relative sensitivity as a function of dose rate for a 6 MV x ray beam. The sensitivities are normalized to their average value for all dose rates. The lowest dose rate of 15 cGy/min was achieved underneath a MLC. A dose rate of 16.5 cGy/min (star) was achieved at an SSD of 424.5 cm. Within $\pm 2.5\%$, the dosimeter's sensitivity shows dose rate independence from 15 to 1000 cGy/min. The error bars represent 1 standard deviation of repeated measurements.

FIG. 9. The relative sensitivity of KCl:Eu²⁺ dosimeters as a function of field size and depth. The results were normalized to the calibration conditions, i.e., at a depth of 10 cm in a 10×10 cm² field. As the field size and depth increase, the dosimeters' sensitivities increase due to the increase in the scatter-to-primary ratio. As a result, a maximum over-response of 12% was observed in the largest field and depth (25×25 cm² and 20 cm, respectively) used in this study. This represents a significant reduction from a value of 82% observed (data not shown) for an Agfa MD10 CR plate at the same conditions. The error bars represent 1 standard deviation of repeated measurements.

III.B.5. Field size and depth dependence

The effective atomic number for KCl:Eu²⁺ material is 18 (since the amount of europium is of the order of ppm it has a negligible contribution to the effective Z). Due to the Z^3 dependence of the photoelectric mass attenuation coefficient, an over-response to low-energy scattered photons was expected for KCl:Eu²⁺ dosimeters compared to ideal water-equivalent dosimeters. This was evaluated by examining the dosimeters' responses to variations in field size and depth. These data are indicative of the effect of using a single dose response calibration curve for measurements with multiple effective field sizes and multiple depths. As the field size and depth increase, the scatter-to-primary dose ratio increases, so detectors that are oversensitive to low-energy photons will overestimate the dose at the field center. Figure 9 shows the relative sensitivities of KCl:Eu²⁺ dosimeters that received a 200 cGy dose (6 MV x ray) at various field sizes and depths. The results were normalized to the calibration conditions, i.e., at a depth of 10 cm in a 10×10 cm² field. As the field size and depth increased, the dosimeters' sensitivities increased due to the increase in the scatter-to-primary ratio. As a result, a maximum over-response of 12% was observed in the largest field and depth (25×25 cm² and 20 cm, respectively) used in this study. As a comparison, EDR2 film exhibited less energy dependence (8%), although it had a larger effective Z of 43. For the commercially available BaFBr_{0.85}I_{0.15}:Eu²⁺-based CR plate, as much as 82% over-response was observed (data not shown).

III.B.6. Field size and depth dependence with low-energy photon filters

As reported by several authors,^{5,21,24,25,29} the over-response of a dosimeter could be reduced by placing low-energy photon filters in the solid water phantom sandwiching

FIG. 10. The relative sensitivity of KCl:Eu²⁺ dosimeters as a function of field size and depth in the presence of low-energy photon filters. 0.3 mm lead foils sandwiching the dosimeters remove the over-response at large fields and depths. The dosimeters showed weak depth dependence with an average change of 3.7%. The error bars represent 1 standard deviation of repeated measurements.

the dosimeter. As shown in Fig. 10, the dosimeters showed weak depth dependence with an average change of 3.7%.

III.B.7. Angular dependence

Figure 11 shows the angular dependence of the dosimeter's sensitivity with 0° defined as the dosimeter's surface normal to the beam axis. The sensitivity shows weak dependence at small angles with a 5% drop up to 45°. A value of as large as 20% was observed for 90° when the sensitive surface of the dosimeter was parallel to the beam axis. This angle dependence is probably due to the cylindrical shape of the dosimeter, which is inherently anisotropic.

IV. DISCUSSION

Contemporary practices in radiation oncology require multidimensional quantitative dose delivery validation. Although radiographic film has been the standard medium for imaging 2D dose distributions, there are significant drawbacks to this convention. As clinics continue to move toward

FIG. 11. The relative sensitivity of KCl:Eu²⁺ dosimeters as a function of incident beam angle. The sensitivity shows weak dependence at small angles with a 5% drop up to 45°. A value of as large as 20% was observed for 90° when the sensitive surface of the dosimeter was parallel to the beam axis. The error bars represent 1 standard deviation of repeated measurements.

digital imaging, the availability of film and film processors is becoming increasingly limited. In addition, because film is intrinsically a single use detector, it cannot be calibrated reliably and its dosimetric properties cannot be monitored without the assumption that individual films have the same response as others within the batch. CR devices potentially rectify these issues. Indeed, Olch²¹ reported that a CR system performed satisfactory relative and absolute dosimetry for fields less than $15 \times 15 \text{ cm}^2$ when lead filtration was utilized. However, response was enhanced for larger fields likely because the BaFBr_{0.85}I_{0.15}:Eu²⁺ material is highly sensitive to the beam quality. The data shown in this work demonstrate that KCl:Eu²⁺ has a potential to be more conducive to megavoltage dosimetry than BaFBr_{0.85}I_{0.15}:Eu²⁺.

One of the most attractive characteristics of this novel material is its excellent radiation hardness. The crystal lattice of the dosimeter stayed nearly intact after an irradiation to 200 Gy (Fig. 5), indicating that this material could be reused at least 100 times if 2 Gy per use was assumed, for example, for a patient-specific IMRT QA. Improvement of the radiation hardness is expected. A recent work,³⁰ for example, investigated the influence of lithium (Li) codoping on the radiation hardness of CsBr:Eu²⁺. It was reported that the integration of Li⁺ into the CsBr:Eu²⁺ lattice suppressed the generation of *M* centers during x ray irradiation and thereby partially improved the radiation resistance. It should also be noted that the loss in sensitivity that was observed in the present study is clinically manageable as calibration of the dosimeter to known dose during the same dose measurement session is a standard QA practice.

As shown in Fig. 4, the PSL signal faded with time after x ray irradiation and eventually reached a fading rate of about 0.1%/hour after 12 h. This indicates that the dosimeter should be read the day after x ray irradiation to avoid having to apply a time-dependent correction. The relatively fast fading during the first few hours might be due to shallower electron or hole traps in the forbidden band, which is consistent with the findings that the PSL signal remains almost the same for readout at wait times of 20, 44, and 68 h after correcting for fading and readout signal depletion. Further research would involve the identification of these energy levels using thermoluminescence or photoluminescence techniques in the infrared region. If shorter wait time is desired, these low-energy level traps could be, for example, optically emptied by using a low-power infrared stimulation light during a preread cycle.

A KCl:Eu²⁺ dosimeter has considerably lower effective *Z* (*Z*=18) than commercially available CR material. Thus, a reduction in over-response to low-energy scattered photons for a KCl:Eu²⁺ dosimeter compared with a BaFBr_{0.85}I_{0.15}:Eu²⁺ dosimeter was expected. However, XV or EDR2 demonstrated a lower over-response to low-energy photons than KCl:Eu²⁺ even though AgBr has much higher effective *Z* of 43. This is likely due to the much thinner active layer of the radiographic film, which is 0.2 μm (Ref. 5) compared to 1 mm for the KCl:Eu²⁺ dosimeter used in this study. Energy dependence in a dosimeter is largely the

result of secondary electrons created through low-energy scattered photon interactions within the sensitive volume. Palm *et al.*¹³ found that the fraction of photons in the spectrum with energy below 100 keV was correlated with measured radiographic film dose artifacts. The projected maximum range for, say, a 50 keV secondary electron generated by photoelectric interactions in water is about 40 μm .²⁸ Because the range in a medium is roughly independent of *Z* for electrons (and positrons),²⁸ the above range in AgBr reduces to about 6 μm when scaled by AgBr's physical density of 6.473 g/cm³. This is much larger than the thickness of the sensitive layer of radiographic film. Thus it is highly probable for secondary electrons to deposit a larger fraction of their energy outside of the dosimeter's sensitive volume. If this reasoning holds true, it is expected that a thin dosimeter made of KCl:Eu²⁺, with a thickness of much less than the range of low-energy secondary electrons in the material, would provide significantly lower energy dependence than radiographic film and would have an overall dose response close to water. Modern thin film techniques, for example, tape casting³¹ or vacuum evaporation,^{32,33} are capable of producing dosimeters in such thin layers. Note that our experiments (data not shown) demonstrate that current KCl:Eu²⁺ dosimeters are easily able to detect sub-cGy dose using, for example, a 5 mW He-Ne yellow laser. We expect that further materials processing optimization, including synthesis conditions, particle morphology, doping uniformity, etc., will result in materials that provide sufficient signal strength over a clinically relevant dose range for a KCl:Eu²⁺ dosimeter with thickness of the order of 10 μm .³⁴ This effort will be supported in part by the intrinsic low background noise associated with storage phosphor techniques.^{23,35,36}

One advantage of having a reusable dosimeter is that characteristic curves can be predetermined and a library of curves stored. For example, if a dosimeter is used for a specific clinical case, the appropriate sensitometric curve could be selected from a stored library. When the sensitometric curves are acquired in similar beam and phantom geometries as the measurement dosimeter, the effect of the moderately high *Z* on the PSL energy dependence of a KCl:Eu²⁺ dosimeter may become clinically insignificant. More importantly, because the dosimeter is reusable, the energy dependence artifact could be reliably rectified using Monte Carlo techniques developed by other authors.^{13,15,16} These approaches would become more practical with the future clinical use of Monte Carlo treatment planning.

V. CONCLUSIONS

The work reports the quantitative use of KCl:Eu²⁺ dosimeters for radiation therapy dosimetry. (1) KCl:Eu²⁺ dosimeters exhibited supralinear response to dose after irradiation by a linear accelerator from 0 to 800 cGy. There were no measureable changes in the stimulation spectra within this dose range. (2) KCl:Eu²⁺ showed satisfactory radiation hardness. There was no significant change in the stimulation spectra after irradiation to 100 and 200 Gy compared to a fresh dosimeter. This indicates that this material could be

reused at least 100 times if 2 Gy per use was assumed, for example, for patient-specific IMRT QA. (3) After x ray irradiation, the PSL signal faded with time and eventually reached a fading rate of about 0.1% /h after 12 h. This indicates that the dosimeter should be read at least 12 h postirradiation if the lowest uncertainty is desired. (4) There was no dependence on x ray dose rate ranging from 15 to 1000 cGy/min. (5) The over-response to low-energy photons was comparable to radiographic film, for example, a Kodak EDR2 film. Reduction in dosimeter thickness to a few microns is hypothesized to reduce the energy dependence.

Future work involves the optimization of materials processing in order to further improve the dosimetric characteristics, for which a better understanding of, for example, energy transfer, sensitivity, radiation hardness, and signal fading mechanisms will be essential. The successful fabrication of a planar panel using modern thin film techniques could provide a reusable, high-resolution two-dimensional dosimeter with dose response close to water.

ACKNOWLEDGMENT

This work was supported in part by NIH Grant No. R21CA131690.

^{a)} Author to whom correspondence should be addressed. Electronic mail: hli@radonc.wustl.edu

¹IMRTCWG, "Intensity Modulated Radiation Therapy Collaborative Working Group: Intensity-modulated radiotherapy: Current status and issues of interest," *Int. J. Radiat. Oncol., Biol., Phys.* **51**, 880–914 (2001).

²S. A. Leibel, Z. Fuks, M. J. Zelefsky, S. L. Wolden, K. E. Rosenzweig, K. M. Alekhtiar, M. A. Hunt, E. D. Yorke, L. X. Hong, H. I. Amols, C. M. Burman, A. Jackson, G. S. Mageras, T. LoSasso, L. Happersett, S. V. Spirou, C. S. Chui, and C. C. Ling, "Intensity-modulated radiotherapy," *Cancer J.* **8**, 164–176 (2002).

³G. A. Ezzell, J. M. Galvin, D. Low, J. R. Palta, I. Rosen, M. B. Sharpe, P. Xia, Y. Xiao, L. Xing, and C. X. Yu, "Guidance document on delivery, treatment planning, and clinical implementation of IMRT: Report of the IMRT Subcommittee of the AAPM Radiation Therapy Committee," *Med. Phys.* **30**, 2089–2115 (2003).

⁴J. M. Galvin, G. Ezzell, A. Eisbrauch, C. Yu, B. Butler, Y. Xiao, I. Rosen, J. Rosenman, M. Sharpe, L. Xing, P. Xia, T. Lomax, D. A. Low, and J. Palta, "Implementing IMRT in clinical practice: A joint document of the American Society for Therapeutic Radiology and Oncology and the American Association of Physicists in Medicine," *Int. J. Radiat. Oncol., Biol., Phys.* **58**, 1616–1634 (2004).

⁵S. Pai, I. J. Das, J. F. Dempsey, K. L. Lam, T. J. LoSasso, A. J. Olch, J. R. Palta, L. E. Reinstein, D. Ritt, and E. E. Wilcox, "TG-69: Radiographic film for megavoltage beam dosimetry," *Med. Phys.* **34**, 2228–2258 (2007).

⁶J. L. Robar and B. G. Clark, "The use of radiographic film for linear accelerator stereotactic radiosurgical dosimetry," *Med. Phys.* **26**, 2144–2150 (1999).

⁷I. J. Chetty and P. M. Charland, "Investigation of Kodak extended dose range (EDR) film for megavoltage photon beam dosimetry," *Phys. Med. Biol.* **47**, 3629–3641 (2002).

⁸N. Dogan, L. B. Leybovich, and A. Sethi, "Comparative evaluation of Kodak EDR2 and XV2 films for verification of intensity modulated radiation therapy," *Phys. Med. Biol.* **47**, 4121–4130 (2002).

⁹J. Esthappen, S. Mutic, W. B. Harms, J. F. Dempsey, and D. A. Low, "Dosimetry of therapeutic photon beams using an extended dose range film," *Med. Phys.* **29**, 2438–2445 (2002).

¹⁰A. J. Olch, "Dosimetric performance of an enhanced dose range radiographic film for intensity-modulated radiation therapy quality assurance," *Med. Phys.* **29**, 2159–2168 (2002).

¹¹X. R. Zhu, P. A. Jursinic, D. F. Grimm, F. Lopez, J. J. Rownd, and M. T. Gillin, "Evaluation of Kodak EDR2 film for dose verification of intensity modulated radiation therapy delivered by a static multileaf collimator," *Med. Phys.* **29**, 1687–1692 (2002).

¹²P. M. Charland, I. J. Chetty, S. Yokoyama, and B. A. Fraass, "Dosimetric comparison of extended dose range film with ionization measurements in water and lung equivalent heterogeneous media exposed to megavoltage photons," *J. Appl. Clin. Med. Phys.* **4**, 25–39 (2003).

¹³A. Palm, A. S. Kirov, and T. LoSasso, "Predicting energy response of radiographic film in a 6 MV x-ray beam using Monte Carlo calculated fluence spectra and absorbed dose," *Med. Phys.* **31**, 3168–3178 (2004).

¹⁴N. L. Childress, M. Salehpour, L. Dong, C. Bloch, R. A. White, and I. I. Rosen II, "Dosimetric accuracy of Kodak EDR2 film for IMRT verifications," *Med. Phys.* **32**, 539–548 (2005).

¹⁵A. Palm and T. LoSasso, "Influence of phantom material and phantom size on radiographic film response in therapy photon beams," *Med. Phys.* **32**, 2434–2442 (2005).

¹⁶A. S. Kirov, G. Caravelli, A. Palm, C. Chui, and T. LoSasso, "Pencil beam approach for correcting the energy dependence artifact in film dosimetry for IMRT verification," *Med. Phys.* **33**, 3690–3699 (2006).

¹⁷M. J. Butson, T. Cheung, and P. K. Yu, "Weak energy dependence of EBT gafchromic film dose response in the 50 kVp–10 MVp x-ray range," *Appl. Radiat. Isot.* **64**, 60–62 (2006).

¹⁸O. A. Zeidan, S. A. Stephenson, S. L. Meeks, T. H. Wagner, T. R. Willoughby, P. A. Kupelian, and K. M. Langen, "Characterization and use of EBT radiochromic film for IMRT dose verification," *Med. Phys.* **33**, 4064–4072 (2006).

¹⁹B. D. Lynch, J. Kozelka, M. K. Ranade, J. G. Li, W. E. Simon, and J. F. Dempsey, "Important considerations for radiochromic film dosimetry with flatbed CCD scanners and EBT GAFCHROMIC® film," *Med. Phys.* **33**, 4551–4556 (2006).

²⁰E. Wilcox, G. Daskalov, and L. Nedialkova, "Comparison of the Epson expression 1680 flatbed and the Vidar VXR-16 dosimetry PRO film scanners for use in IMRT dosimetry using gafchromic and radiographic film," *Med. Phys.* **34**, 41–48 (2007).

²¹A. J. Olch, "Evaluation of a computed radiography system for megavoltage photon beam dosimetry," *Med. Phys.* **32**, 2987–2999 (2005).

²²H. Nanto, K. Murayama, T. Usuda, F. Endo, Y. Hirai, S. Taniguchi, and N. Takeuchi, "Laser-stimulable transparent KCl:Eu crystals for erasable and rewritable optical memory utilizing photostimulated luminescence," *J. Appl. Phys.* **74**, 1445–1447 (1993).

²³J. A. Rowlands, "The physics of computed radiography," *Phys. Med. Biol.* **47**, R123–R166 (2002).

²⁴S. E. Burch, K. J. Kearfott, J. H. Trueblood, W. C. Sheils, J. I. Yeo, and C. K. Wang, "A new approach to film dosimetry for high energy photon beams: lateral scatter filtering," *Med. Phys.* **24**, 775–783 (1997).

²⁵I. J. Yeo, A. Beiki-Ardakani, Y. B. Cho, M. Heydarian, T. Zhang, and M. Islam, "EDR2 film dosimetry for IMRT verification using low-energy photon filters," *Med. Phys.* **31**, 1960–1963 (2004).

²⁶H. Nanto, F. Endo, Y. Hirai, S. Nasu, S. Taniguchi, and N. Takeuchi, "Eu-doped KCl phosphor crystals as a storage material for two-dimensional ultraviolet-ray or x-ray imaging sensors," *J. Appl. Phys.* **75**, 7493–7497 (1994).

²⁷J. Zimmermann, S. Hesse, H. v. Seggern, M. Fuchs, and W. Küpfer, "Radiation hardness of CsBr:Eu²⁺," *J. Lumin.* **114**, 24–30 (2005).

²⁸F. H. Attix, *Introduction to Radiological Physics and Radiation Dosimetry* (Wiley, New York, 1986).

²⁹I. J. Yeo, C. K. Wang, and S. E. Burch, "A filtration method for improving film dosimetry in photon radiation therapy," *Med. Phys.* **24**, 1943–1953 (1997).

³⁰J. Zimmermann, S. Hesse, H. v. Seggern, M. Fuchs, and W. Knüpfen, "Influence of Li-codoping on the radiation hardness of CsBr:Eu²⁺," *J. Appl. Phys.* **101**, 113711 (2007).

- ³¹H. Li, E. Epelbaum, M. Batentschuk, and A. Winnacker, "Tape casting of storage phosphor BaFBr:Eu²⁺ for x-ray imaging," *Mater. Sci. Eng., B* **96**, 313–319 (2002).
- ³²P. Leblans, L. Struye, and P. Willems, "A new needle-crystalline computed radiography detector," *J. Digit Imaging* **13**, 117–120 (2000).
- ³³B. Schmitt, M. Fuchs, E. Hell, W. Knüpfer, P. Hackenschmied, and A. Winnacker, "Structured alkali halides for medical applications," *Nucl. Instrum. Methods Phys. Res. B* **191**, 800–804 (2002).
- ³⁴Y. Zheng, Z. Han, J. P. Driewer, D. A. Low, and H. Li, "Prefabrication Monte Carlo study of a storage phosphor film having nearly water equivalent dose response" (unpublished).
- ³⁵Y. Amemiya and J. Miyahara, "Imaging plate illuminates many fields," *Nature (London)* **336**, 89–90 (1988).
- ³⁶H. Li, A. L. Gonzalez, H. Ji, and D. M. Duggan, "Dose response of BaFBrI:Eu²⁺ storage phosphor plates exposed to megavoltage photon beams," *Med. Phys.* **34**, 103–111 (2007).