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A bench-top megavoltage fan-beam CT using CdWO$_4$-photodiode detectors. I. System description and detector characterization

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We describe the components of a bench-top megavoltage computed tomography (MVCT) scanner that uses an 80-element detector array consisting of CdWO$_4$ scintillators coupled to photodiodes. Each CdWO$_4$ crystal is $2.75 \times 8 \times 10$ mm$^3$. The detailed design of the detector array, timing control, and multiplexer are presented. The detectors show a linear response to dose (dose rate was varied by changing the source to detector distance) with a correlation coefficient ($R^2$) nearly unity with the standard deviation of signal at each dose being less than 0.25%. The attenuation of a 6 MV beam by solid water measured by this detector array indicates a small, yet significant spectral hardening that needs to be corrected before image reconstruction. The presampled modulation transfer function is strongly affected by the detector’s large pitch and a large improvement can be obtained by reducing the detector pitch. The measured detective quantum efficiency at zero spatial frequency is 18.8% for 6 MV photons which will reduce the dose to the patient in MVCT applications. The detector shows a less than a 2% reduction in response for a dose of 24.5 Gy accumulated in 2 h; however, the lost response is recovered on the following day. A complete recovery can be assumed within the experimental uncertainty (standard deviation <0.5%); however, any smaller permanent damage could not be assessed. © 2006 American Association of Physicists in Medicine. [DOI: 10.1118/1.2181290]

Key words: megavoltage computed tomography, scintillators, detective quantum efficiency, radiation damage

I. INTRODUCTION

Thick, segmented scintillation detectors have the potential for low-dose computerized tomographic (CT) imaging of the patient, while using the same beam generation system as is used for the treatment. There have been several approaches to obtain CT images of the patients prior to treatment.\(^1\)-\(^{12}\) One group has placed a diagnostic CT scanner on rails,\(^1\) where the patient couch is rotated by a half-turn between imaging and treatment. While the image quality of a diagnostic CT scanner is unparalleled, this system does incur a much higher cost due to the larger treatment room as well as the purchase and regular maintenance of a diagnostic CT scanner. Cone beam CT scanners have been implemented on the gantry of a linear accelerator.\(^2\) The image quality of the kilovoltage cone beam CT (CBCT) has been shown to be adequate for the purpose of patient position verification; however, this system requires an x-ray tube and an additional flat panel imager. Although a megavoltage CBCT (Refs. 3–9) has poor image quality and a relatively large radiation dose, it may be a simple and inexpensive solution for CT imaging in the treatment position. Megavoltage CT (MVCT) uses the same beam generation system as that used for treatment. The detector system used for MVCT can also be utilized for portal imaging and dose reconstruction.\(^10\) MVCT images do not suffer from artifacts due to metallic objects found in dental fillings, prosthesis, and brachytherapy applicators; although postprocessing techniques have been developed to reduce the metallic artifacts in kVCT.\(^11\) MVCT numbers are linearly related to electron density,\(^12\) where as the kVCT numbers deviate from a single linear curve for high atomic number objects. Although a piece-wise linear curve is generally used...
for kVCT calibration\textsuperscript{13} to partially compensate for this behavior, the dense bone kVCT numbers can vary up to 6.4% depending on the location and the size of the phantom used.\textsuperscript{12} The performance of a CT scanner is measured in terms of three basic parameters: High contrast spatial resolution, soft tissue contrast (low contrast details), and radiation dose delivered to the patient.\textsuperscript{14} High resolution is required to verify patient position with CT images, but adequate soft tissue contrast is required to differentiate tumors embedded in normal tissue. It is also important that the radiation dose delivered to the patient in verification CT studies be kept low lest the accumulated dose for the entire treatment exceed the critical tissue tolerance. In comparison to diagnostic scanners, at megavoltage energies the dose delivered to the patient per interaction is high\textsuperscript{15} and the detective quantum efficiency (DQE) of the detectors is low.\textsuperscript{16,17} In order to improve the DQE, thick scintillator-photodiode detectors have been used previously in fan-beam MVCT,\textsuperscript{18-20} but the basic detector characteristics were not measured. Nakagawa \textit{et al.}\textsuperscript{21} used cadmium tungstate (CdWO\textsubscript{4}) with photodiodes in fan-beam MVCT; however, their CdWO\textsubscript{4} (<5 mm) was too thin to achieve a high DQE. Keller \textit{et al.}\textsuperscript{22} performed a detailed Monte Carlo simulation of a single-row curved detector based on pressurized xenon gas and demonstrated that the large DQE of the focused detector (20.4% in a 4 MV photon beam) was mainly due to the tungsten septa plates. This detector was used in a fan-beam MVCT (Ref. 15) and it was also utilized for dose reconstruction.\textsuperscript{23}

Recently, our laboratory published\textsuperscript{24} the results of modeling scintillation-photodiode detectors at MV energies and showed that an array of 10 mm thick CdWO\textsubscript{4} scintillation crystals coupled to silicon photodiodes provided 19% and 26% DQE in 6 MV and 1.25 MeV beams at zero frequency, respectively. The fundamental advantages of CdWO\textsubscript{4} scintillation crystals and photodiodes are as follows: CdWO\textsubscript{4} is a dense scintillator (7.9 g/cm\textsuperscript{3}) giving a large quantum efficiency for a small thickness. The optical yield in modern CdWO\textsubscript{4} is large enough so that the DQE is determined mainly by the x-ray quantum and energy deposition noises. Moreover, the spatial spread of signal in detector arrays made with a reflective coating on individual elements can be reduced. Importantly for CT, the afterglow of CdWO\textsubscript{4} is only 0.05% at 3 ms.\textsuperscript{24} The radiation damage in CdWO\textsubscript{4} is very small.\textsuperscript{25,26}

We have fabricated a one-dimensional (1D) array of 80 elements containing CdWO\textsubscript{4} and photodiodes, which are placed on an arc with a radius of 110 cm. A rotary stage was added to create a small third-generation CT scanner. This paper presents our design of detector electronics, data acquisition timing control, detector electronics, analog multiplexer, precision rotary stage, and data acquisition board. Objects to be imaged are rotated by a precision rotating stage (200RT, Daedal Division, Parker Hannifin Corp, Irwin, PA) that is driven by a stepper motor (ZETA57-83, Compumotor Division, Parker Hannifin Corp, Rohnet Park, CA). The direction and speed of rotation are controlled by the digital port and counter timer on a data acquisition board’s (NI PCI-6034E, National Instruments, Austin, TX). The data acquisition board also contains a 16-bit analog-to-digital (A-to-D) converter used for digitizing the detector data. The user interface to the data collection, display, and storage are implemented in LABVIEW graphical programming language (National Instruments, Austin, TX).

\textbf{II. MATERIALS AND METHODS}

\textbf{II.A. System description}

The block diagram of the prototype MVCT system is shown in Fig. 1 along with arrows indicating the control signals and data flow between five subsystems: Data acquisition timing control, detector electronics, analog multiplexer, precision rotary stage, and data acquisition board. The active area of each photodiode element is 1.175 \times 2 mm\textsuperscript{2}; each crystal (2.75 \times 8 mm\textsuperscript{2} cross section) illuminates two consecutive photodiode elements as shown in Fig. 2. The components of these detector boards have been described earlier\textsuperscript{27} and the bond-
ing process of the eight-element CdWO$_4$ crystal array has
been discussed. Briefly, eight $2.75 \times 8 \times 10 \text{ mm}^3$ CdWO$_4$
crystals are bonded together in white gelcoat (Ashland
Chemical Type 1 polyester gelcoat), commonly used as the
outer coat for fiberglass boats. The crystals were spaced
0.4 mm apart, which is the same as the dead space in-
between the photodiode elements. The same distance was
also maintained between the ten consecutive detector boards
while placing the boards on an arc with a radius of 110 cm,
so that the detector pitch is 3.15 mm throughout the
80-element array.

II.A.2. Detector electronics and multiplexing unit

Although one can purchase an off-the-shelf integrated cir-
cuit that contains an integrator, sample-and-hold, and A-to-D
converter (e.g. DDC114, TI, Dallas, TX), its internal capac-
tor is too small to collect the large charge produced by
a single 6 MV pulse. Therefore, the detector electronics were
designed using discrete components. The electronics for a
single detector channel have been described earlier, and
for this work printed circuit boards each containing eight-
detector channels have been fabricated. The electronics of a
single detector channel are shown in Fig. 3. Contrary to the
circuit design in Ref. 27, the circuit in Fig. 3 does not contain
a 10 nF capacitor in the feedback loop of the amplifier to
increase the data collection speed beyond 339 Hz. We expect
the resulting increase in electronic noise to be negligible
compared to the signals produced by irradiation of the crys-
tals. Also, the first stage of multiplexing (8:1) that occurs on
each eight-element detector board is not shown in Fig. 3.
The photocurrent from two photodiode elements is integrated by
a gated integrator that uses a 1 nF capacitor. The analog
switch in the feedback loop of the integrator controls the
charging and discharging of the capacitor through an $\bar{I}/D$
control signal where $\bar{I}$ and $D$ stand for integrate and dis-
charge, respectively. Since the output per pulse for the 6 MV
beam is significantly larger than the instantaneous dose rate of
the Co$^{60}$ beam, the gain of the output amplifier was set to
47 for the Co$^{60}$ and unity for the 6 MV experiments.

II.A.3. Data acquisition timing control (DATC)

The timing diagram for the various signals is shown in
Fig. 4. The DATC generates timing control signals for the
detector electronics ($\bar{I}/D$ and $\bar{S}/H$), 80:1 analog multiplexer
(a seven-bit binary number for detector selection), and
A-to-D conversion clock (“Detector read” in Fig. 1). Data
acquisition is initiated by a home signal generated by a mag-
etic switch on the rotary stage. The rate of data acquisition
is determined by a trigger signal that is the “Sync” signal for
the 6 MV linear accelerator ($T_{\text{CYCLE}}=7.3$ ms) or is derived
from an independent clock for Co$^{60}$ ($T_{\text{CYCLE}}=1.44$ ms).
The low-voltage Sync signal is available at the front of the con-
trol console in all Varian linear accelerators outside the treat-
ment room. This signal was connected to our “data acquisi-
tion timing control” system through an optoisolator circuit to
reduce interference noise. Immediately following the trigger,$\bar{I}/D$ is asserted for $T_{\text{INT}}=0.8$ ms to integrate the photocur-
cent. The $\bar{S}/H$ ($\bar{S}=$sample; $H=$hold) signal is asserted to
hold the integrated signal just prior to discharging the capaci-
tors in the gated integrator i.e., $T_{\text{SAMPLE}}=0.7$ ms. The inte-
grated signals for all the detector channels are held for
$T_{\text{HOLD}}=0.64$ ms for Co$^{60}$ and 4.0 ms for 6 MV, which
allows the A-to-D converter to read the data for all 80 channels
at a rate of $f_{A\rightarrow D}=125$ kHz for Co$^{60}$ or 20 kHz for 6 MV.
During acquisition, the seven-bit binary number is also in-
cremented synchronously with $f_{A\rightarrow D}$ to select the multi-
plexed output for detectors 1 through 80.

Since the data collection is synchronized with sync pulses
from the linear accelerator, each data point represents the
radiation detected per pulse. In general, the dose rate control
mechanism of the linear accelerator decides whether or not a

Fig. 3. The detector electronics for a single element consists of a gated
integrator ($C_{\text{int}}=1 \text{ nF}$), an amplifier (gain=$R_2/R_1$) and a sample-and-hold
circuit.

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given Sync pulse will produce radiation. For example, with a setting of 250 monitor units (MU) per minute, five out of six Sync pulses on average are associated with a radiation pulse. Therefore, the first step in processing the data is to discard the nonradiation pulses. This step is not required for the continuous Co\textsuperscript{60} radiation case. For all of the measurements, the average signal of each channel in the absence of radiation, i.e., the dark signal, is subtracted from the collected data.

**II.B. Detector characterization**

The system has run reliably except for an initial problem with the detector boards. It was noticed very early in the experimental stage that the analog switches on the detector boards are sensitive to leakage radiation through the accelerator jaws. Therefore, we built extra shielding blocks to protect the detector board electronics. It should also be noted that the detector boards have a practical limitation in that the change in the detector gain required between the 6 MV and Co\textsuperscript{60} experiments was obtained by physically replacing the resistances in the feedback loop of the amplifiers. Therefore, experiments could not easily be conducted in both the 6 MV and 0.6 MV settings of 250 monitor units MU. Most of the experiments in this work were thus performed in a 6 MV photon beam.

To characterize the detector, we measured the detector linearity with respect to dose rate; attenuation of a 6 MV beam by solid water; the presampled MTF, NPS, DQE, and the reduction in the detector signal as a function of the radiation dose accumulated in the detector.\textsuperscript{16,28,29}

**II.B.1. Detector response to dose rate**

This experiment was designed to measure the linearity of the detector with respect to dose rate in a 30.0 × 2.2 cm\textsuperscript{2} 6 MV photon beam. In order to vary the dose rate, the source to detector distance (SDD) was varied from 100 cm to 150 cm in 5 cm steps. At each SDD, 2000 data points were collected per detector channel. The mean signal for each detector was calculated after discarding the nonradiation pulses. The mean detector signal as a function of (100/SDD)\textsuperscript{2} was subjected to linear regression.

**II.B.2. Attenuation measurement**

The fundamental requirement for a CT detector is to measure the attenuation produced by the scanned object in an accurate and linear manner. Thus, we used our detector array to measure the attenuation of a 6 MV photon beam by solid water. Slabs of solid water (12 slabs each 2 cm thick) were placed on top of the treatment couch in the path of a narrowly collimated 6 MV photon beam (25×2.2 cm\textsuperscript{2}) while the detector was placed under the treatment couch at a SDD of 110 cm. The solid water was placed on the tennis racket part of the couch that has negligible attenuation of the 6 MV beam. For each thickness of solid water, one thousand readings per detector channel were acquired and the means calculated. The entire experiment was repeated five times. The attenuation factor for each solid water thickness was calculated using the following: Attenuation=Ln (nonattenuated signal/attenuated signal). A second-order polynomial fit was used to study the effects of spectral hardening. A straight line was fit to the first four data points and extrapolated to greater solid water thickness while a second-order polynomial was fit to all data points. If these measurements were carried out in a broad beam, the scattered radiation created in the solid water phantom would introduce nonlinearity in attenuation measurements. We have carried out these measurements using a narrow beam (25×2.2 cm\textsuperscript{2}) to protect the electronic hardware from radiation damage.

**II.B.3. Presampled MTF**

As discussed in Ref. 28, the shift invariance of segmented scintillator detectors is maintained for the shifts that are integer multiples of their pitch as long as the crystals and photodiodes are registered; this is true for this detector since one crystal covers exactly two photodiodes. Therefore, the classical Fourier techniques can be used for the detector system analysis. The MTF measured previously\textsuperscript{24} for the eight-element prototype array suffered from significant aliasing due to the large detector pitch of \(x_0=3.15\) mm. Therefore, the presampled line spread function (LSF) and the resulting MTF for the 80-element array was measured using the theory described by Fujita et al.\textsuperscript{30} With our approach, the LSF was over sampled by a factor of 5 to minimize aliasing. The measurement procedure is depicted in Fig. 5. A slit beam was centered on one detector element (indicated by coordinate 0 in Fig. 5) and the average signal for each detector element for 10,000 pulses was obtained. The detector array was then translated to four other positions indicated by \(−2δ, −δ, δ,\) and \(2δ (δ=x_0/5)\) with respect to the slit beam location, and

![Diagram](https://via.placeholder.com/150)

**Fig. 4.** The timing diagram of the control signals: I/D=Integrate-discharge control of the gated integrator; S/H=sample-and-hold control; Detector Read=convert clock for the A-to-D converter. The trigger signal is either the "Sync" signal of the pulsed radiation (6 MV) or derived from an independent clock for continuous radiation (Co\textsuperscript{60}).
the data acquisition was repeated. The resulting data were interlaced to give the presampled LSF, which was Fourier transformed to obtain the presampled MTF.

The experimental setup for measuring the presampled LSF is shown in Fig. 6. The long lead blocks are placed on their own stage to facilitate the collimation of a slit beam. Extra lead blocks near the source are used to reduce the leakage radiation from the secondary jaws of the accelerator. The slit beam of 0.2 mm width was collimated (accelerator field size=4×4 cm²) using two 25 cm×5 cm×10 cm lead blocks separated by a 0.2 mm thick aluminum shims. The width of the slit was verified by taping an XV film at the detector end of the lead blocks and irradiating for 100 MU. The film was then scanned (Vxr-16, Vidar Systems Corporation, Herndon, VA) and the full width at half maximum of the slit image was determined to be 0.25 mm. Initially, the slit was approximately placed at the center of Detector Element “36” and then the translation stage was moved in very small increments until the measured (and corrected) signals in Detector Elements “35” and “37” were the same; this was taken as the zero position of the slit on Detector Element 36. The measured data were divided by the mean signal for each detector obtained in open beam measurements to correct for the element-to-element sensitivity variations. The detector array was moved in 0.63 mm increments by a manually operated translation stage. The entire experiment was repeated six times to determine the experimental uncertainty.

By using a theoretical derivation similar to Fujita et al., it can be shown that the measured presampled MTF is the product of the radiation MTF, which characterizes the signal spread due to charge particle transport, Compton scattering and optical leakage within the crystal array, and the detector aperture MTF. Importantly, the aliasing in the presampled MTF, dominated by the detector aperture, is insignificant since the detector aperture MTF is very small for spatial frequencies larger than five times the Nyquist frequency of the detector array, i.e., 0.8 line pairs per mm (lp/mm).

The presampled MTF obtained from this procedure was divided by the detector aperture MTF in order to obtain the radiation MTF. The detector aperture MTF results from a combination of the photodiode and crystal size effects. As shown in Fig. 2, a single crystal (2.75 mm) sits on two photodiodes (2×1.175 mm) with a 0.4 mm dead reflector gap in-between. Thus, one cannot take crystal size as the detector aperture because there is no direct photodetector in 0.4 mm reflector gap. Also, one cannot just take the aperture size of 2×1.175=2.35 mm as the reflector sends some light back into crystal; a fraction of this reflected light may actually fall back on the active photodetector part and contribute to signal. The effective detector aperture was estimated as (1/fo) where fo is the first zero-crossing spatial frequency in the measured presampled MTF. The detector aperture MTF is determined as a sinc function using (1/fo) as the effective aperture width.

To ensure that the measured radiation MTF did not suffer from aliasing, a Monte Carlo calculation of the radiation MTF was performed. Using the XYZDOSnrc user code of the EGSnrc Monte Carlo transport code,13 we modeled a slit beam (0.2 mm×8 mm) of 6 MV photons incident on the center of an 8 mm wide, 30 mm long, and 10 mm deep CdWO₄ crystal array. The slit beam was incident on the 8 mm×30 mm face of the crystal array, which was divided into 0.01 mm×8 mm×10 mm voxels to reduce aliasing in the simulated LSF. The x-ray transport simulation in CdWO₄ crystals was also performed in our previous work, however, the lateral voxel dimension was 3.15 mm instead of 0.01 mm used in this work. The spectrum of the 6 MV photons was the same as used by Lachaine et al.32 A total of 1 million photon histories were run using the transport parameters AE=E_cut=0.521 MeV and AP=P_cut=0.01 MeV. The average energy deposited into chosen voxels of the CdWO₄ crystal block was calculated and taken as the radiation LSF, which was Fourier transformed to give the radiation MTF. The standard deviation in the calculated LSF was less than 3% up to 10 mm distance from the slit beam. It should be mentioned that our EGSnrc (Ref. 31) Monte Carlo calculation includes the scattered radiation produced within the detector array.
II.B.4. Noise power spectrum

Ideally, the noise-only data can be obtained by subtracting the detector data that are collected between two consecutive pulses. However, the pulse-to-pulse variation of the 6 MV beam prevented an accurate measurement of the NPS as noticed previously. Therefore, a periodogram average method was used. This particular detector is a long narrow strip in which the longer dimension is segmented with a pitch of \( x_0 \) (3.15 mm) and the detector has a finite sensitive width \( L \) perpendicular to the direction of array. Therefore, the NPS can be measured using the synthesized slit technique reviewed by Williams et al. It should be noted that the integration at each pixel over the slit width \( L \) occurs naturally in this case. Based on the discussion of the synthetic slit technique by Williams et al. and using Eq. (15) from that reference, the NPS of the slit, \( \text{NPS}_{NS} \), is related to the two-dimensional \( \text{NPS}_{2D} \) as follows:

\[
\text{NPS}_{NS}(u) = \int \text{NPS}_{2D}(u,v)|T(v)|^2 dv,
\]

where \( u \) and \( v \) are the spatial frequency coordinates, and \( T(v) \) is the Fourier transform of the response function of the detector along the width \( L \) denoted as \( R(y) \). In a synthesized slit technique, this response function is a rectangular function of width \( L \). However, in the present detector, the width of the crystals is 8 mm, while the width of the photodiodes is 2 mm (Fig. 2). The photodiodes receive scattered optical photons from the crystal area beyond 2 mm width, as well as some of the optical photons that fall on the nonsensitive part of the photodiode array are reflected back into the crystals; a portion of which is again detected within the sensitive 2 mm width. Therefore, the response function of the detector in the width dimension is not rectangular and it is represented as such in Eq. (1). According to Williams et al., if the width \( L \) is large enough such that \( \text{NPS}_{2D} \) has a small change in \( u \) direction over the narrow width of \( T(v) \), then one obtains the following approximation between \( \text{NPS}_{NS} \) and \( \text{NPS}_{2D} \):

\[
\text{NPS}_{NS}(u) \approx \text{NPS}_{2D}(u,0) \int |T(v)|^2 dv.
\]

Therefore, the NPS along the array dimension of the present detector is \( \text{NPS}_{2D}(u,0) \) and evaluated from the natural slit NPS by dividing it by the integral in Eq. (2). The \( \text{NPS}_{NS} \) of sampled random data \( \Delta S \) of length \( N \) from our detector is given by Eq. (16) in Ref. 33 as follows:

\[
\text{NPS}_{NS}(u) = \frac{x_0}{N} (|\text{DFT}(\Delta S)|^2),
\]

where \( N \) is the number of samples used in the discrete Fourier transform (DFT) and notation \( (\cdot) \) represents the expected value and,

\[
\text{NPS}(u) = \text{NPS}_{2D}(u,0) \approx \frac{x_0}{N\int|T(v)|^2 dv} (|\text{DFT}(\Delta S)|^2).
\]

Since the integral in the denominator of Eq. (4) has units of \( \text{mm}^{-1} \), the NPS resulting from equation has units of \( \text{volt}^2 \text{mm}^{-2} \).

The detector was centered in a 2.5 \( \times \) 32 cm\(^2\) 6 MV field and the data for 10,000 pulses were recorded at a SDD of 110 cm. We have used a narrow beam in the NPS measurements to protect the nearby electronic components from radiation damage. This irradiation geometry does not include any small change in the measured NPS caused by the increased head scatter in the broad clinical beams. The resulting data were divided by the mean signal of each detector element to remove the effects of beam profile and element-to-element sensitivity variations. For each radiation pulse, \( i \), these normalized data, \( S_{i} \), were divided into five subgroups, each containing 16 detector elements. For each of the five subgroups, \( k \), the noise only data, \( \Delta S_{i,k} \), for each radiation pulse were obtained by subtracting the first eight elements from the second eight elements. This step is necessary in order to avoid the large correlation among detector elements caused by the pulse-to-pulse fluctuations in the beam output. A DFT of the resulting eight-point noise \( (N=8)\)-only data was performed to obtain one periodogram. Finally, the NPS was obtained by averaging these periodograms over the five subgroups and total number of radiation producing pulses, \( I \), as follows:

\[
\Delta S_{i,k}(1:8) = S_{i}(16k - 15:16k - 8) - S_{i}(16k - 7:16k)
\]

\[
1 \leq k \leq 5, \quad 1 \leq i \leq I,
\]

\[
\text{NPS}(u) \approx \frac{x_0}{(2 \cdot 5 \cdot I \cdot N \cdot \int |T(v)|^2 dv)} 
\times \sum_{i=1}^{I} \left( \sum_{k=1}^{5} |\text{DFT}(\Delta S_{i,k})|^2 \right),
\]

\[
u = \frac{n}{8 \cdot x_0}(\text{lp/mm}); \quad n = 0,1,2,3,4.
\]

Please note that the summations and division by \( (5I) \) approximate the expected value by an average value. The NPS was divided by 2 in Eq. (6) since \( \Delta S_{i,k} \) is obtained by subtraction. The entire experiment was repeated ten times to assess the experimental errors. As a result of division by the mean signal, the aforementioned procedure calculates the normalized NPS in units of \( \text{mm}^{-2} \).

The detector response function \( R(y) \) was measured to evaluate the integral in Eq. (6) as follows. A slit beam of 0.4 mm width was formed by the same apparatus as used in the MTF measurements. The detector array was arranged such that the length of the slit was along the array dimension and the width along the width \( L \) of the crystals. The entire detector array was translated along the crystal width and past the slit width in steps of \( \Delta y \) (0.375 mm), and the detector data was measured for 5000 pulses at each step. The mean detector signal, \( D(n\Delta y) \), as a function of slit position relative to the center of crystal width, normalized at center, represents the response function.
\[ R(n\Delta y) = \frac{1}{L} D(n\Delta y); \quad \text{for} \ n = 1, N_y \text{ and } N_y = \frac{L}{\Delta y}, \] (8)

The integral in the denominator of Eq. (6) was then evaluated as follows:

\[
\int |T(u)|^2 \, dv = \Delta v \sum_{j=0}^{N_y-1} \text{DFT}(R(n\Delta y))^2; \\
\text{and } \Delta v = \frac{1}{\Delta y N_y} \text{ mm}^{-1}. \] (9)

**II.B.5. Detecte quantum efficiency**

The DQE is calculated from the measured MTF and NPS as follows:34

\[
\text{DQE}(u) = \frac{\text{MTF}^2(u)}{\text{NPS}(u)/\bar{S}^2\varphi}, \] (10)

where \( \varphi \) is the photon fluence (quanta per mm\(^2\)) impinging on the detector and \( \bar{S} \) is the average signal detector. The average signal \( \bar{S} \) is unity since the detector data were normalized to remove the signal variations. The photon fluence impinging on the detector per radiation pulse, \( \varphi \), was obtained by using the following method. The dose per unit fluence factor \( (F = 7.51 \times 10^{-8} \text{ cGy mm}^2/\text{photon}) \) was calculated by Lachaine et al.32 in water who used the EGSnrs Monte Carlo software with the 6 MV spectrum of this machine. Rogers35 calculated the dose per unit fluence factors for a number of monoenergetic photons in ICRU four-element tissue by using broad beam geometry simulations and the previous version of Monte Carlo simulation software (EGS3). Using the factors in Ref. 35 in Tables 9–12 at 1.5 cm depth, and the relative fluence spectrum of the 6 MV beam for our machine, we estimated the dose to fluence factor to be 7.94 \( \times 10^{-8} \text{ cGy mm}^2/\text{photon} \). Since the monoenergetic factors were not calculated in water exactly the same geometry as the calibration condition of the linac, and the resulting polyenergetic factor is only 5.7% different from that calculated by Lachaine et al.32 we used a value of 7.51 \( \times 10^{-8} \text{ cGy mm}^2/\text{photon} \) in our calculations. Therefore,

\[
\varphi = \frac{\dot{D}}{F}. \] (11)

The dose per radiation pulse \( \dot{D} \) was calculated as follows. The treatment beam was calibrated to deliver 1 cGy per MU at 1.5 cm depth in water for a \( 10 \times 10 \text{ cm}^2 \) field at 100 cm from the source. Using this geometry a pinpoint chamber (N31006, PTW Freiburg) was placed at 1.5 cm depth in a solid water phantom. The chamber reading \( (r_{ch}) \) was recorded as 100 MUs were delivered. A second reading \( (r) \) was obtained using a 2.5 cm \( \times 32 \text{ cm} \) field size at a distance of 110 cm from source to replicate the irradiation condition of the NPS measurement. Therefore, the dose delivered at 1.5 cm depth in the geometry of NPS measurements was obtained as \( 100 \frac{r}{r_{ch}} \) cGy. The number of pulses, \( N \), that actually produced radiation was determined from the data acquired with the detector array. Therefore, \( \dot{D} \) is given as \( (100 r/r_{ch}) \).

In order to further establish that the fluence calculation is reasonable, we calculated the zero-frequency DQE of the detector using the method suggested by Swank36 and used by Keller et al.22 for a 1D xenon gas based arc detector array. The absorbed energy distribution \( (\text{AED} = E = n\Delta E) \), is the probability that an incident photon from the 6 MV spectrum deposits \( E \) amount of energy in the detector array. Monte Carlo simulation, similar to that used in the radiation MTF calculation, was also used to score the AED \( \text{AED}(E = n\Delta E) \) in the entire crystal, i.e., by ignoring voxels, in 0.01 MeV \( \Delta E \) energy bins. The zero-frequency DQE \( (0) \) was calculated as \( M_2^2/M_2 \), where \( M_1 \) and \( M_2 \) are, respectively, the first- and second-order energy moments of the AED defined as follows:

\[
M_j = \sum_n (n\Delta E)^j \text{AED}(n\Delta E); \quad j = 1, 2 \] (12)

where \( n \) denotes that energy bin number and the summation is taken over 600 energy bins.

**II.B.6. Radiation damage**

Radiation hardness of CdWO\(_4\) crystals has been studied previously25,26 in a Co\(^{60}\) beam for doses of \( 10^5 \) Gy and \( 10^6 \) Gy. Since CdWO\(_4\) scintillation crystals have been used extensively in single slice diagnostic CT systems, the radiation damage to the crystal-photodiode pair is not a significant issue if utilized for MVCT alone. However, the final goal of this project is to create a two-dimensional (2D) detector array that can be utilized both for imaging and exit fluence measurements during radiotherapy treatments. In this situation, the detector array may receive up to 30 Gy per day based on 50% transmission through 30 patients each receiving a dose fraction of 2 Gy. Thus, we measured the effect of approximately 25 Gy per day on the detector array.

To avoid damaging the electronic components with the leakage radiation from the linear accelerator, the experiment was carried out in the Co\(^{60}\) beam using a narrow beam collimation. The detector array was placed 84.5 cm from the source in a cobalt teletherapy unit (780E, Therantronics, Kanata, Canada). The collimator of the Co\(^{60}\) unit was set at 4.2 \( \times 7.8 \text{ cm}^2 \) and the irradiation field was further collimated to about 0.8 \( \times 8.0 \text{ cm}^2 \) by two lead blocks. The beam was turned on for 1.2 min and an average of 5000 readings for each detector element was collected at the end of the radiation period; the total time of data acquisition was 0.25 s. The detector was given a 5 min recovery period to simulate the minimum time between two consecutive patient treatments. The 1.2 minute irradiation and subsequent 5000 point data collection was repeated. The cycle of irradiation and recovery period was repeated 20 times. Using the calibration data and the output factor for a 4.2 \( \times 7.8 \text{ cm}^2 \) field, each irradiation of 1.2 min is estimated to deliver a dose of 1.36 Gy to a small mass of tissue in free space. This value is multiplied by the tissue air ratio, estimated as 1.0075 for Co\(^{60}\), of the collimated field at 0.5 cm depth, and ratio of the mass absorp-
tion coefficient of CdWO₄ to tissue (0.895) to give 1.23 Gy to CdWO₄ for each 1.2 min irradiation and amounting to a total dose of 24.5 Gy. The mass absorption coefficients of tissue and CdWO₄ are available from NIST.²⁷ The experiment was repeated on four different days. The mean and standard deviation of the mean detector signal over four days were calculated and analyzed as a function of accumulated radiation dose. CdWO₄ is not expected to significantly recover its lost sensitivity during the irradiation period because of the long (10–20 h) recovery period,²⁵ thus, the measured radiation damage is not expected to have a dose rate effect. Therefore, the radiation damage in a 6 MV pulsed beam should be similar to that in Co⁶⁰ beam for a given accumulated dose to CdWO₄.

III. RESULTS

III.A. Detector response to dose rate

The linear regression analysis performed on the detector signal as a function of (100/SDD)² indicated a very linear response of the detector with dose rate. The R² coefficient was better than 0.9998 for all elements in the detector array. Since the standard deviation for each data point is less than 0.25%, the data points follow a straight line very well. However, there was a considerable (> two times) variation in the slopes indicating a large element-to-element variation in the sensitivity.

III.B. Attenuation measurement

The attenuation of 6 MV photons by solid water is plotted in Fig. 7 for detector elements 1 and 40. Detector element 40 was positioned close to the central axis of the beam and Detector Element 1 was located 12.4 cm off axis. Therefore, Detector Element 1 saw a slightly larger solid water thickness due to the diverging path. The deviation of the second-order fit from the straight line fit to the first four points for Element 1 at larger solid water thickness indicates the presence of spectral hardening. At a solid water thickness of 20 cm, the measured attenuation is about 5.5% lower than predicted by the extrapolated straight line. The second-order polynomial described the measured attenuation data (standard deviation of <1%) very well for all the detector elements (R² = 1.0). For Detector Element 1, the second-order component amounts to 7.5% of the total attenuation at a thickness of 20 cm. For Detector Element 40, the measured attenuation of the beam is less than that for Detector Element 1. This is due to a conical-shaped flattening filter in the linac beam causing the photon beam to be more penetrating on the central axis.

III.C. Presampled MTF

The measured and calculated MTFs are plotted in Fig. 8. Since the measured presampled LSF spanned the length of the array (80×3.15=252 mm), the resolution of the measured presampled MTF in frequency space is 0.003968 mm⁻¹ i.e., 1/(252 mm) that gives measured points close to zero frequency. For reference, the Nyquist frequency of the detector array is only 0.16 lp/mm due to the large pitch. The radiation MTF was obtained by dividing the presampled MTF by the aperture MTF; it was not calculated in the vicinity of zero crossings of the aperture MTF to avoid magnification of errors. Aliasing in the measurement of the presampled MTF is negligible up to the Nyquist frequency of the detector array because the presampled MTF is small at 0.8 lp/mm. This is also suggested by the MTF from the Monte Carlo simulation (also shown in Fig. 8) which is higher than the radiation MTF at all spatial frequencies. The Monte Carlo calculated radiation MTF can be compared with measured radiation MTF in the low spatial frequency range. The Monte Carlo calculated MTF is determined by the charged particle transport and, to a lesser extent, by Compton scattering.

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Fig. 7. Plot of the attenuation of 6 MV photons by solid water as measured by Detector Elements 1 and 40. The straight line fit to the first four points of the Detector Element 1 data is extrapolated to increased thickness to indicate the amount of spectral hardening. Second-order polynomials fit to all data describe the spectral hardening very well. The attenuation data for the remaining elements lies in between Elements 1 and 40. The error bars on the measured data points are smaller than the symbols.

Fig. 8. The presampled aperture and radiation MTFs of the detector array. For reference, the Nyquist frequency of the detector array is 0.16 lp/mm. The radiation MTF is obtained by dividing the presampled MTF by the aperture MTF. The error bars on the crystal MTF are larger due to the division. Radiation MTF-MC is calculated using Monte Carlo simulation.
III.D. NPS

The measured relative response of the detector, \( R(n\Delta y) \), in perpendicular to the array direction is shown in Fig. 9 as a function of the distance from the center of crystals. This function is used to calculate the integral in Eq. (9) as part of NPS normalization in Eq. (6). Although the crystal width is 8 mm, the detectors provide signals even when the slit is located beyond 4 mm from the center because of the additional reflective coating at these surfaces. Charged particles created in coating materials travel to the crystals and create signal. The sensitive detector width, \( L \), in Eq. (8) was determined to be 10 mm as the extent of the non-zero signal in Fig. 9. The measured NPS of the detector array up to the Nyquist frequency is plotted in Fig. 10.

III.E. DQE

The measured presampled and radiation MTFs were used to calculate the DQE of the detector array, and the results are plotted in Fig. 11. The DQE at zero spatial frequency is about 18.8% and decreases at higher spatial frequencies. The higher DQE using the radiation MTF shows that the large detector pitch of 3.15 mm adversely affects the DQE at higher spatial frequencies. In a future implementation of this detector, a smaller element pitch will be considered to reduce aliasing effects on the MTF and DQE. The value of 18.8% for the measured DQE is in good agreement with the 19% calculated previously using a two-step Monte Carlo simulation. The first- and second-order energy moments of the AED determined by the Monte Carlo simulation in the present work were calculated to be 0.35 and 0.65 respectively that also resulted in zero-frequency DQE of 19% using Swank’s method. This calculation indirectly verifies that the fluence per pulse estimate and the NPS normalization are reasonable.

III.F. Radiation damage

Plotted in Fig. 12 is the mean detector signal as a function of the accumulated dose to scintillation crystals. Although,
only three detector elements are shown, the data for the other detector elements showed similar trends. The error bars show ±1 standard deviation (<0.5%) of the mean detector signals over four experiments. The data clearly show that the mean detector signal decreases continuously as a function of accumulated dose; however, the reduction in signal is less than 2% for all detector elements. The data also indicate that a significant portion of the small reduction in mean signal is recovered by the next day. However, because the experimental set up needed to be dismantled at the end of each day the measurements varied randomly due to the uncertainty in positioning the detector in the beam.

As a result of this uncertainty, we did not observe a consistent increase or decrease in the detector data from day-to-day. Figure 12 also includes the minimum and maximum values for detector element 42. The curve corresponding to the minimum values was observed on the first day of irradiation while the curve corresponding to the maximum values was observed on the third day of irradiation. Since the temperature coefficient of the optical yield in CdWO₄ is less than 0.1%/°C (at 25 °C), small day-to-day variation in the room temperature (<1 °C) is not likely to create inconsistency in measured data.

It should also be noted that these data cannot be easily compared with previous studies because they looked at the reduction in luminescence at the wavelength of peak emission due to very large (10³–10⁶ Gy) dose in much larger CdWO₄ crystals.

IV. DISCUSSION

There is a need for improving the DQE of MVCT detectors in order to image the patient in treatment position with a low dose. High density thick scintillators such as CdWO₄ have the potential to provide the improvement in DQE of MVCT detectors. We have performed several experiments with the CdWO₄ detector array to determine its suitability for MVCT imaging.

The data in the attenuation measurement may suggest that a correction for spectral hardening is required for the fan-beam data acquired by the prototype MVCT system. It may also suggest that such corrections vary from element to element due to the presence of conical flattening filter. It should be noted that the element-to-element correction (usually referred as the flood field correction) removes the effects of element-to-element sensitivity variation and the variation in primary fluence across the detector. Since it is based on open-beam measurement, it does not account for the variation in the attenuation by imaged objects due to the variation in the primary beam spectrum across the detector.

The difference between the measured radiation MTF and the Monte Carlo calculated MTF could potentially be caused by the leakage radiation through and scattered radiation produced within the 25 cm long lead blocks. However, the same blocks were also used in measuring the MTF of similar detector in Ref. 24 where a two-step Monte Carlo approach was also used to include the effect of optical photon transport. A very good agreement between the measured MTF and that obtained from two step Monte Carlo approach was obtained. Since two-step Monte Carlo approach did not account for the leakage and scattered radiation within the lead blocks, this agreement indirectly suggested that leakage and scattered radiation in lead blocks is negligible. The measured radiation MTF has extra spreading of signal due to the leakage of optical photon through the optical glue and small optical photon leakage through the reflective gelcoat. Most of the leakage occurs through the common optical glue sheet placed between the bottom surface of crystal arrays and top surface of the photodiode arrays. Therefore, the measured radiation MTF is significantly lower than the Monte Carlo calculated radiation MTF. Since the radiation MTF is obtained by deconvolution, it has large errors at frequencies larger than 0.24 lp/mm. Even within the lower-frequency range, the aperture MTF dominates the shape of the presampled MTF. Since the presampled MTF is dominated by the aperture MTF at low frequencies, decreasing detector pitch should improve the presampled MTF. However, the spatial resolution in the projection data is also determined by the focal spot size of the linear accelerator and geometric magnification. Therefore, an optimal detector pitch should be determined by taking all these factors, including the radiation and optical transport in scintillators, into account which is part of a future investigation. Since the detector array is similar in configuration to the xenon gas arc detector described by Keller et al., the measured radiation MTF can be compared with the MTF presented in Fig. 12 of Ref. 22 for a tungsten plate thickness of 0.32 mm corresponding to the arc detector used in their bench top MVCT. The two MTFs are comparable up to 0.2 lp/mm and CdWO₄ MTF (0.46) is larger than xenon gas detector MTF (0.3) around 0.5 lp/mm. Additionally, the frequencies at 0.5 MTF ($f_{50}$) are around 0.43 and 0.32 lp/mm, respectively, for
CdWO₄ and xenon gas detector MTFs, although the large error bars on CdWO₄ MTF should be kept in mind in making the comparison.

The relative decrease in the measured NPS as a function of spatial frequency is generally compared with the squared magnitude of the unsampled MTF to understand the noise correlation caused by the x-ray absorption noise. For the case of present detector with a large pitch, such a comparison would indicate that the DQE does not decrease as rapidly with frequency as the unsampled presampled MTF due to aliasing in the measured NPS. This overestimation of the measured NPS, caused by aliasing, at nonzero spatial frequencies will tend to underestimate the measured DQE at nonzero frequencies.

Imaging arc detector using tungsten plates immersed in pressurized xenon gas has been studied by Keller et al. A Monte Carlo simulation performed on a 1D array by Keller et al. calculated DQEs of 20.4% and 31.4% in a 4 MV beam, respectively, for the focused and nonfocused arrangements corresponding to 0.32 mm tungsten plate thickness. However, the nonfocused detector may suffer from poor spatial resolution. The zero-frequency DQE of CdWO₄ array is about 19% in a 6 MV and it is expected to be larger in 4 MV beam. The advantage of our CdWO₄ array is that the DQE can be further increased by using slightly thicker crystals if the optical isolation between detector elements is maintained. Moreover, back-illuminated photodiode arrays can be tiled in 2D because the wire bonding and optical windows of the photodiodes are on the opposite sides of the substrate. These arrays will allow a focused, high DQE, 2D detector to be built. The loss of resolution due to divergent rays in unfocused detectors was studied by Moyle and alternative schemes, using flat panel photodiode arrays, of focusing thick scintillation crystals to source were suggested by Pang and Rowlands and Sawant. We are investigating the use of 2D tiled photodiode arrays to build focused MVCT detectors.

The radiation damage experiments show that the CdWO₄-photodiode detector will show a reduction in signal of <2.0% if the detector is used to measure the exit fluence for every patient on a particular day. Also, a significant portion of the lost sensitivity will be recovered by the next day within the experimental uncertainty of <0.5%. However, any permanent loss (or gain) in the detector sensitivity of less than 0.5% can neither be established nor ruled out using these data because we did not observe a consistent increase or decrease in detector data from day-to-day due to setup uncertainty. A better controlled experimental will be required to measure any permanent damage to this detector due to large doses of radiation. The loss of sensitivity caused by radiation damage is unlikely to produce ghosting artifacts. Phantom artifacts have been described by Siewerdsen et al. for an indirect detection active matrix flat panel detector due to signal lag, and in direct detection active matrix flat panel detectors due to space charge and recovery processes have time constant of the order of reading cycle of the detector. In the imaging experiments, the dose delivered to the CdWO₄ array is expected to be less than 10 cGy. Since the reduction in signal for approximately 25 Gy accumulated to the detector is less than 2%, the effect of 10 cGy would be negligibly small to create any ghosting effect. It can be argued that the detector can suffer a slight sensitivity loss if it is left in the beam during patient treatment and, thus, may receive larger doses. However, since there is negligible radiation damage during the imaging experiment and the recovery time constant for lost sensitivity is of the order of several hours, the radiation damage phenomenon is unlikely to cause any ghosting effect. The radiation damage due to small dose per pulse ( <0.05 cGy) is not expected to create ghosting in pulse-to-pulse data collection. Also, synchronization of data collection with the linac pulse interval of 7.3 ms will not create significant lag because of the small afterglow (0.05% at 3 ms).

V. CONCLUSIONS

We have fabricated a prototype detector array of CdWO₄ photodiodes to be used in a bench-top fan-beam MVCT scanner. The detector electronics, data multiplexer, data acquisition timing control, rotary stage control, and the data acquisition system for this prototype have been designed and shown to work well. The detector responds linearly to dose rate. A small yet significant amount of spectral hardening effect was determined in the attenuation measurements of 6 MV photon beam by solid water. This will form the basis of spectral hardening correction used in processing of the fan-beam projection data in subsequent work.

The presampled MTF of this detector is dominated by the large pitch of 3.15 mm and the radiation MTF shows that a reduction in detector pitch may have a sizeable improvement on the spatial resolution. The most important feature of this detector is the large DQE(0) of 18.8%. In its own class of detectors, i.e., scintillation photodiodes, the detector offers the largest DQE per unit crystal thickness and suggests a significant reduction in radiation dose in MVCT imaging experiment. This issue is even more important since the CdWO₄-photodiode array can be extended into 2D aided by the recent advent of 2D backilluminated photodiode arrays and advances in array fabrication.

Radiation damage to CdWO₄-photodiode combination is a significant issue; however, no previous data exist that concern the radiation therapy irradiation pattern. The experiments conducted in this manuscript indicate that a reduction in detector sensitivity of less than 2% can be expected over a one day period and the detector recovers a large fraction of its lost sensitivity overnight. Any permanent damage to the detector array cannot be established.

In summary, a CdWO₄-photodiode detector array offers a large DQE. In subsequent work, we will investigate the imaging performance of this detector in MVCT application.

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