A silicon strip detector dose magnifying glass for IMRT dosimetry

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Purpose: Intensity modulated radiation therapy (IMRT) allows the delivery of escalated radiation dose to tumor while sparing adjacent critical organs. In doing so, IMRT plans tend to incorporate steep dose gradients at interfaces between the target and the organs at risk. Current quality assurance (QA) verification tools such as 2D diode arrays, are limited by their spatial resolution and conventional films are nonreal time. In this article, the authors describe a novel silicon strip detector (CMRP DMG) of high spatial resolution (200 μm) suitable for measuring the high dose gradients in an IMRT delivery.

Methods: A full characterization of the detector was performed, including dose per pulse effect, percent depth dose comparison with Farmer ion chamber measurements, stem effect, dose linearity, uniformity, energy response, angular response, and penumbra measurements. They also present the application of the CMRP DMG in the dosimetric verification of a clinical IMRT plan.

Results: The detector response changed by 23% for a 390-fold change in the dose per pulse. A correction function is derived to correct for this effect. The strip detector depth dose curve agrees with the Farmer ion chamber within 0.8%. The stem effect was negligible (0.2%). The dose linearity was excellent for the dose range of 3–300 cGy. A uniformity correction method is described to correct for variations in the individual detector pixel responses. The detector showed an over-response relative to tissue dose at lower photon energies with the maximum dose response at 75 kVp nominal photon energy. Penumbra studies using a Varian Clinac 21EX at 1.5 and 10.0 cm depths were measured to be 2.77 and 3.94 mm for the secondary collimators, 3.52 and 5.60 mm for the multileaf collimator rounded leaf ends, respectively. Point doses measured with the strip detector were compared to doses measured with EBT film and doses predicted by the Philips Pinnacle treatment planning system. The differences were 1.1% ± 1.8% and 1.0% ± 1.6%, respectively. They demonstrated the high temporal resolution capability of the detector readout system, which will allow one to investigate the temporal dose pattern of IMRT and volumetric modulated arc therapy (VMAT) deliveries.

Conclusions: The CMRP silicon strip detector dose magnifying glass interfaced to a TERA ASIC DAQ system has high spatial and temporal resolution. It is a novel and valuable tool for QA in IMRT dose delivery and for VMAT dose delivery. © 2010 American Association of Physicists in Medicine. [DOI: 10.1118/1.3264176]

Key words: silicon strip detector, intensity modulation, quality assurance, dose magnifying glass (DMG), dosimetry

I. INTRODUCTION
Intensity modulated radiation therapy (IMRT) enables the delivery of escalated radiation dose to tumor while sparing the critical organs close to the irradiated target. In doing so, IMRT plans tend to produce steep dose falloff regions at the interface of the target and the organ at risk. Currently, quality assurance (QA) verification of IMRT plans is patient based and the techniques commonly used are based on point dose...
measurement using ion chamber and planar fluence verification using films, electronic portal imaging devices (EPIDs), and two dimensional (2D) diode or ion chamber arrays. The ion chamber, though highly reliable, tends to overestimate the penumbra width due to its larger volume, therefore may not accurately represent areas with high dose gradient falloff. The use of films involves calibration of the film and film scanners which delays the QA process. The use of EPIDs as a QA tool require complicated conversion of image to dose, whereas the 2D and 3D diode and ion chamber arrays currently are limited by their spatial resolution in high dose gradient regions. Notwithstanding their limitations, these existing techniques are widely used in many facilities. In order to measure high gradient dose regions generated by IMRT plans, one needs to have a detector with a high spatial resolution. Hence, the sensitive volume of the detector has to be small in physical size. In this paper, we present the concept of a novel detector with high spatial resolution suitable for measuring steep dose gradient areas such as those formed in an IMRT plan. The silicon strip detector, designed and produced by the Centre for Medical Radiation Physics (CMRP), University of Wollongong, is comprised of an array of 128 phosphor implanted $n^+$ strips on a $p$-type silicon wafer having a spatial resolution of 0.2 mm. This detector system is able to look at radiation profiles at submillimeter spatial resolution and subsecond temporal resolution; hence, we called it the “dose magnifying glass” (DMG). We present the application of such a detector in the dosimetric verification of an IMRT plan, comparing it to the planned dose predicted by the radiation therapy planning system. Prior to using this detector in the IMRT plan, a full characterization of the detector was performed including dose per pulse dependency, depth dose curve, stem effect, linearity, uniformity, energy response, angular response dependency, and penumbra width study.

II. MATERIAL AND METHODS

Experience gained in detector technology from high energy physics experimental research on tracking vertex detectors and data acquisition (DAQ) systems has transferred to medical physics over the past decade. This experience was mostly implemented in nuclear medicine imaging systems (PET and gamma cameras), utilizing pixelated semiconductor detectors and coupled with multichannels readout electronics (Medipix). Implementation of such detectors in radiation therapy is slower and further opportunities in this area still exist. Since 2003 CMRP had been developing the concept of a silicon high spatial resolution dosimetry system with high dynamic range (of the order of $10^5$) for applications on medical linacs. The system is based on silicon strip detector technology. The DAQ system is an important issue in high spatial resolution real time dosimetry due to requirements for simultaneous readout of multiple channels (sensors). The prototype system of a Si strip detector with 128 channels was designed and two TERA 64 channel chips were proposed for the instantaneous readout of each strip.

II.A. TERA chip-DAQ system description

The TERA chip is a very large scale integration application specific integration circuit (VLSI ASIC) that was designed by Istituto Nazionale di Fisica Nucleare (INFN)-Torino Division and University of Torino microelectronics group working on the readout of pixelated ionization or strip chambers for hadron therapy. It underwent several modifications during the past decade. The ASIC is based on a current to frequency converter, followed by a digital counter. In this way, the charge can be measured by counting the number of pulses generated by the converter in a given time. The conversion is based on the recycling integrator principle. Each ASIC has 64 independent channels. Each counter is followed by a 16 bit register, with a common load command. It is therefore possible to store the content of the counters at a given time for intermediate readout during treatment. The output of the 64 registers is multiplexed over a single 16 bit output bus. The multiplexer is controlled by an external 6 bit address bus. The output bus has CMOS output drivers and can be put in the tristate mode. It is therefore possible to connect more than one chip on the same bus on the readout board. A schematic diagram of TERA ASIC is presented in Fig. 1. The TERA ASIC chip uses a charge balancing, zero dead time, integration technique to obtain a dynamic range in excess of $10^5$ with a nonlinearity of less than 1%. In this case, the ASIC is capable of measuring currents from a few pA to a few $\mu$A by integrating the current on a 600 fF capacitor via an operational transconductance amplifier (OTA). The integrated current, or charge, is reduced by a fixed amount $Q_c$ via a charge subtraction circuit (with a “subtraction” capacitance, $C_{sub}=200$ fF) when the OTA output voltage ($V_{OTA}$) reaches a user defined threshold level ($V_{th}$). The subtraction charge is defined by $Q_{sub}=C_{sub}V_{sub}$, where $V_{sub}$ is user defined. This charge subtraction method is far superior to traditional resetting of the integrating capacitor since during the charge subtraction the circuit is still active, i.e., the input current continues to be integrated, and therefore there is no associated dead time. The accuracy of the comparator circuit defines the minimum equivalent $Q_c$ (100 fC in our case), and hence the lower limit of the dynamic range. The upper limit of the dynamic range is limited by the maximum voltage of the comparator circuit (the supply voltage: 5 V),

![Fig. 1. Schematic diagram of TERA ASIC (Ref. 57).]
Various versions of the TERA family of ASICs have been successfully implemented in different 2D and 3D “Magic Cube” ionization chambers in hadron therapy. The latest version (Tera 6.0) of the chip was used by Scanditronix-Wellhöfer (IBA group) in commercial dosimeters used in radiotherapy (MR TeraMatriXX and StarTrack) based on gas ionization chambers. More information on TERA chip can be found in the Torino Radiotherapy Group website. Work presented in this paper is the first we are aware of that implements the TERA chip for a solid state detector dosimetry system.

II.B. The CMRP dose magnifying glass (CMRP DMG)

Silicon diodes are widely used for dosimetry in radiation therapy. Similar to ion chamber dosimetry, it is based on measurements of current or charge, which are proportional to the dose rate or accumulated dose, respectively. They have good spatial resolution, usually of the order of 1 mm, and a long life span for relative dosimetry. The advantage of Si detectors for radiotherapy on a linac is in the constancy of the silicon-water electron stopping power ratio over a wide energy range. The detailed operational theory of Si dosimetric diodes can be found in Ref. 14. Silicon detectors have some disadvantages due to their dose rate dependence, angular dependence, energy dependence, and radiation damage. Dose rate dependency can be reduced to 2% by using p-Si diodes from low resistivity silicon and preirradiation of diodes to reduce the initial carrier lifetime in the base material. Radiation damage effects can be almost eliminated by preirradiation of the detector in an electron beam and using low resistivity p-type Si diodes which have a much better radiation hardness than n-type Si diodes.15 However, new n-type diodes that are heavily doped with platinum have also shown good performance, comparable to p-type diodes in their tolerance to radiation damage.14 Taking into account that Si diodes are relative dosimeters, they need to be recalibrated periodically.

A planar array of Si diodes with pitch of 1 cm are widely used for QA on linacs for in phantom plan verification,16 e.g., Sun Nuclear MapCHECK (445 Si diodes, 22 x 22 cm² area, 10 mm pitch). Recently, the application of an array of diodes with the size of about 0.78 mm² placed in two orthogonal planes was introduced for IMRT QA (Delta4® and Scanditros®). Both these systems are very helpful; however, the spatial resolution of direct dose mapping is about 5–10 mm.

Pixelated Si detectors were recently proposed and a prototype DAQ system is developed in the framework of the European project MAESTRO.18 Each pixel is based on an $n^+ - p$ junction surrounded by a guard-ring structure implanted on an epitaxial 50 μm thick $p$-type Si layer grown on a Czochralski substrate, a monolithic silicon segmented module. The best compromise between granularity and electronic complexity has been achieved by choosing pixels with 2 x 2 mm² active area and 3 mm pitch. The discrete readout electronics for 441 pixels/channel (21 x 21 pixels) was developed. The system demonstrated good results but is quite
bulky due to discrete readout electronics.\textsuperscript{20,21} The next version will utilize nine Tera 6.0 chips for readout of segmented detectors. The system in development offered better resolution than 2D discrete diodes which have a spatial resolution of 3 mm.

The Si strip detector is another monolithic position sensitive detector, which is based on an array of \( p-n \) junction strips. Silicon strip detectors were used to characterize a Sr-90 intravascular brachytherapy source in terms of dose depth curves in a tissue equivalent material and the homogeneity of the activity.\textsuperscript{22} Recently, the application of 128 channel strip detectors with sensitive area of each strip \( 0.25 \times 0.25 \text{mm}^2 \) and pitch of 0.3 mm was reported for stereotactic radiotherapy on a linac.\textsuperscript{23} The strip detector is based on high resistivity (1–10 k\( \Omega \) cm) \( n \)-Si with \( p^+ \) implanted strips and working under reverse bias. A readout ASIC with a charge sensitive preamplifier in each channel was placed next to each Si strip detector on the same PCB. Pappas \textit{et al.} reported that the system demonstrated good performance in deriving the dose profile of the stereotactic beam on 6 MV linac and the penumbra measurements were in better agreement with the diamond detector measurement.\textsuperscript{23,24}

The CMRP DMG (Fig. 3) comprises of an array of 128 phosphor implanted \( n^+ \) strips on a \( p \)-type silicon wafer designed to work with a TERA chip. The first version of CMRP DMG is mounted on a ceramic substrate. The ceramic substrate was of a higher density as well as minimize dose rate effects. The minority charge carriers’ lifetime \( \tau \) in the boron doped silicon we used is about 170 \( \mu \)s. This corresponds to diffusion length about 740 \( \mu \)m,\textsuperscript{25} which is larger than the silicon wafer thickness of 375 \( \mu \)m. Therefore, good radiation hardness was expected, i.e., no changing of the sensitivity during the irradiation within the useful dose range. The schematic diagram of the strip detector is presented in Fig. 4. Aluminum was evaporated on top of the \( n^+ \) areas. The detector was used in passive mode and readout was carried out with the detector configured as a planar detector with a common electrode \( p^+ \) from the same side as the \( n^+ \) strips.

\section*{II.C. Charge collection in Si strip}

Estimation of the charge collection in a single strip and the corresponding output frequency of pulses from the TERA chip is important. Assume that the linac provides an average dose of 400 cGy/min corresponding to 6.67 eGy/s. This is made up of 200 separate 3–5 \( \mu \)s pulses every second.\textsuperscript{26} This corresponds to 0.33 mGy/pulse. For the estimation purpose, assume that the sensitive volume (100\% charge collection) of the single strip is at least \( 20 \times 5000 \times 300 \mu \text{m}^3 \). Taking the density of Si to be \( 2.33 \times 10^3 \text{ kg/m}^3 \), the mass of Si, \( m \), is \( 6.99 \times 10^{-8} \text{ kg} \). Taking into account that the energy required to produce an electron-hole pair in Si, \( W=3.6 \text{ eV} \) and the dose in Si and water are approximately similar for linac photon energies,\textsuperscript{26,27} we expect

\[ Q = \frac{D \cdot m}{e} \]

\[ = \frac{(3.3 \times 10^{-4} \text{ J}) \times 5.99 \times 10^{-8} \text{ kg}}{3.6 \text{ J/C}} = 6.32 \times 10^{-12} \text{ C}, \]

where \( D \) is the linac dose per pulse, \( m \) is the Si mass, and

\[ W = \frac{3.6 \text{ eV}/e - h \text{ pair} \times 1.602 \times 10^{-19} \text{ J/eV}}{1.602 \times 10^{-19} \text{ C/e}} = 3.6 \text{ J/C}. \]
The total time required to deliver 100 MUs was obtained by CC13 ion chamber measurement with the statistical counts. The actual dose at each measurement position was achieved by irradiating the detector under closed solid collimator jaws at 101.5 cm source to detector distance. For linac measurements, the p-Si strip detector was sandwiched between two small slabs of solid water (2.5×5.0×0.5 cm³) machined to fit the sensitive area (Fig. 5). The solid water is then sandwiched inside another larger piece of PMMA (7.0×15.0×3.0 cm³), which is machined to accept the strip detector inside its solid water encapsulation.

II.D. Dose per pulse measurement

For dose per pulse measurement, the CMRP DMG was mounted on a solid water phantom at 1.5 cm depth and irradiated with 10×10 cm² beam. To obtain the dose per pulse variation, the source to detector distance was varied from 87 to 158 cm. At each position, the detector was irradiated with 6 MV photon beam for 100 MUs (monitor units) at 400 MU/min repetition rate. The lowest linac dose per pulse was achieved by irradiating the detector under closed solid collimator jaws at 101.5 cm source to detector distance. For this setup, 1000 MUs were delivered to obtain sufficient statistical counts. The actual dose at each measurement position was obtained by CC13 ion chamber measurement with the similar setup. The dose pulses delivered per MU by the linac was estimated with the information provided by the vendor manual.28 The total time required to deliver 100 MUs was recorded by the readout system of the CMRP DMG. This was then used to derive the dose per pulse measured by the CC13 and the CMRP DMG.

II.E. Percent depth dose measurement

The strip detector with the PMMA housing was placed on top of a 30×30×15 cm³ solid water block. Sections of solid water and PMMA were used to provide scatter around the detector. The percent depth dose profiles were obtained using solid water at the depths of 1.0–1.5 (at 0.1 cm intervals), 5, 10, 15, and 20 cm for a 10×10 cm² field size, at source to surface distance of 100 cm. Photon energy of 6 MV was used. The result was compared to Farmer ion chamber (NE-2571) measurements in a solid water phantom. The active volume of the Farmer ion chamber has an inner diameter of 6.3 mm and an internal length of 24.0 mm.

II.F. Stem effect measurement

Stem effect is the radiation induced conductivity in the cable or stem of the radiation detector when exposed to radiation beam.29,30 The stem effect of this device was investigated by irradiating the CMRP DMG with a rectangular field size of 20×5 cm² at 1.5 cm depth in solid water phantom. The rectangular field size was chosen so that the narrow width of the field size was sufficient to cover the detector area, while avoiding the cable connectors when the long axis is orthogonal to the cables. Measurements were made with the long axis of the field parallel and perpendicular to the cables. The difference between the two measurements is deemed to be due to the stem effect.

II.G. Linearity measurement

The CMRP DMG was inserted into the I’mRT phantom (IBA Dosimetry)31 and positioned at 10 cm depth. PMMA sections were machined specifically to pack the detector snugly in the I’mRT phantom. The setup was irradiated with 6 MV photons from 5 to 400 MUs delivered with 400 MU/min repetition rate. The corresponding doses delivered to the detector ranged from 3.89 to 311.05 cGy.

II.H. Uniformity measurement

The CMRP DMG consisted of a strip silicon detector with 128 pixels. The response from each detector and its charge amplifier may differ slightly from that of the adjacent detectors. Hence, taking the raw readings from the detector, one would expect slight nonuniformity between the 128 detector/readout channels. A uniformity correction was performed using a 6 MV broad beam irradiation of a 10×10 cm² field size at 1.5 cm depth in phantom. There were also some channels that did not operate due to PCB circuit board manufacturing imperfection. The data from the nonoperational channels were removed prior to analysis. For each experiment, the CMRP DMG is positioned at the central axis of the beam and irradiated. The raw data are first visually inspected and the defective channels are removed from consideration. The remaining channel readings were then averaged to obtain a mean reading, \( \bar{x} \). Then, each individual channel’s reading, \( x_i \), is normalized to \( \bar{x} \), to give the response of the channel and thus creating a correction factor (\( u_i = x_i/\bar{x} \)), assuming flatness of the radiation field. This factor was applied on all the subsequent measurements to correct for the channel-to-channel variation in the detector and amplifier response.

II.I. Energy response measurement

To study the energy response of the CMRP DMG, the strip detector was irradiated under an orthovoltage machine (Guymay DX 3300) which operates at the range of 50–250 kV. The orthovoltage machine was calibrated following the IPEMB 1996 protocol.32,33 The absolute calibra-

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**Fig. 5.** Schematic diagram of the cross section view of CMRP DMG. The silicon detector is mounted on a ceramic substrate and there is a layer of air gap (1.5×33.0 mm²) above the strips.

For \( Q_e = 600 \) fC/count on the TERA,9 the number of counts on the output of TERA will be about 10 counts/linac pulse. That provides approximately 2000 counts/s for quoted dose rate. In the case where charge collection is <100%, the number of measured counts/s may be less. This confirms the suitability of TERA chip for the Si strip detector in linac radiation fields.

For linac measurements, the p-Si strip detector was sandwiched between two small slabs of solid water (2.5×5.0×0.5 cm³) machined to fit the sensitive area (Fig. 5). The solid water is then sandwiched inside another larger piece of PMMA (7.0×15.0×3.0 cm³), which is machined to accept the strip detector inside its solid water encapsulation.

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tion accuracy is 1.9%, whereas the reproducibility of the calibration is better than 0.5%. The detector was irradiated without any buildup. For 50–150 kV tube voltage, a 100 mm diameter circular applicator was used and the detector was placed at a focus to surface distance of 312 mm. For the 200 and 250 kV tube voltage, 100 mm diameter circular applicator was used with detector positioned at focus to surface distance of 512 mm. For high energy photons, the detector was irradiated using the 6 and 10 MV beams from the Varian Clinac 21EX linear accelerator (Varian, Palo Alto, USA) ($d_{\text{max}}$ of 1.5 cm for 6 MV and 2.0 cm for 10 MV). The linac was calibrated according to the IAEA TRS-398 protocol. The relative standard uncertainty of the absolute calibration is approximately 1.7%. 1 Gy water equivalent dose was delivered at all energies. The readings are normalized to 1 at the energy of 6 MV.

II.J. Angular response dependency measurement

Silicon detectors are well known for their directional response dependencies. In addition, the design of the present prototype detector with its ceramic mounting is expected to cause some angular response to the DMG. Hence, we also studied the effect of angular incident beams on CMRP DMG. The strip detector was located in the middle of the I’mRT phantom and positioned at the isocenter. Radiation beams with the field size of $10 \times 10$ cm$^2$ were delivered with static gantry angles at the intervals of 15° from 0° to 90°. The readings were then compared to CC13 ion chamber measurements at the isocenter and at the positions ±1 cm lateral from the isocenter. The angular response is defined as the ratio of the signal measured by CMRP DMG to the ion chamber measurements at each point of measurement. For the CMRP DMG measurement, we have taken the average readings of the detector channels within the distance of ±1.5 mm from the measurement point so as to be comparable to half of the inner diameter of the CC13 ion chamber. The angular response taken at the ±1 cm spatial distance from the center of the detector allowed us to investigate the response of detector channels as a function of distance from the center channel. The angular response correction factor was subsequently used to correct for the data measured in the following IMRT dosimetric plan verification.

II.K. Penumbra measurement

The penumbra of $10 \times 10$ cm$^2$ field size was measured at $d_{\text{max}}$ (1.5 cm) and 10 cm depths in solid water. A reading was first acquired with the detector placed at the central axis of the beam. A second reading was then acquired with the detector placed at the field edge, with the penumbra positioned in the middle of the strip. The readings are then normalized to the average readings measured at the central axis. This measurement technique was necessary because the prototype strip detector is only 2.56 cm wide and is not able to measure the dose at the central axis as well as the field edge, simultaneously. It is envisaged that clinical strip detector systems will incorporate multiple adjacent 128 detector elements, allowing the instantaneous measurement of a full beam profile in a single exposure. Penumbra measurements were obtained for the secondary collimator jaws and the rounded leaf end of the linac multileaf collimator (MLC). The measurements are then compared to measurements in water using Gafchromic EBT (ISP, Wayne, NJ) films in a water tank and solid water phantom.

II.L. Clinical application in IMRT fields

A prostate IMRT plan was selected and copied onto the planning CT images of the I’mRT phantom and recalculated using Pinnacle treatment planning system (TPS) with a dose grid of 2 mm$^3$. The isocenter of the IMRT plan was purposely shifted 4 cm laterally to obtain a high gradient falloff at the position of the detector. The treatment plan was exported to the Aria R & V system. The I’mRT phantom was set up with the strip detector sandwiched in the middle (9 cm in depth). The isocenter was shifted accordingly and treated. Measurements were also taken with Gafchromic EBT film sandwiched at the depth of 9 cm in the I’mRT phantom to be compared with the CMRP DMG measurements. The EBT film was scanned with a flatbed scanner (Epson Perfection V700) after 24 h to allow for postirradiation coloration of the film. The films were scanned at the scanning resolution of 150 dpi in 48-bit RGB mode and analyzed using image analysis software, IMAGE J 1.39U (National Institute of Health, USA). Care was taken to scan the films at the center of the scanner and in the same orientation to avoid scanner-induced nonuniformity and change in the pixel values due to film scanning orientation. Standard calibration films were irradiated in solid water in the same experimental session. The results from the measurements were compared to the Pinnacle predicted dose and EBT film measurements.

III. RESULT

III.A. Dose per pulse

To estimate the dose per pulse, we know that for the Varian Clinac 21EX linear accelerator the gun pulses were delivered at 360 Hz for the 6 MV photon energy. At the repetition rate of 400 MU/min, four out of six gun pulses were beam pulses. Therefore, 240 dose pulses were delivered per second. From our measured data, we know that 100 MU were delivered in approximately 15 s. Hence, 3600 dose pulses were delivered in 100 MUs. The CMRP DMG sensitivity was taken as the total number of counts divided by 3600 dose pulses. The detector response was defined as the sensitivity of CMRP DMG over dose per pulse measured by CC13 ion chamber. Figure 6 shows the detector response over CC13 dose per pulse response normalized to 1 at the source to detector distance of 101.5 cm, corresponding to an estimated dose per pulse of $2.78 \times 10^{-4}$ Gy/pulse. For the range of $3.89 \times 10^{-4}$–$9.45 \times 10^{-5}$ Gy/pulse (corresponding to 87–158 cm source to detector distance), the detector response varied by <5%. For the dose per pulse change of 390-fold, the nonlinearity of the device was found to be 23%. This was obtained from the measurement performed under the closed solid collimator jaws (corresponding to
1.11 × 10^{-6} \text{ Gy/pulse}, the detector under-responded by 23% compared to the CC13 ion chamber. The dose measured at this point is equivalent to the transmission dose of 0.4 cGy.

A correction function was derived to correct for the dose-per-pulse dependence. From Fig. 6, a fourth degree polynomial curve was fitted to the data points. The \( r^2 \) of the curve was 0.994. The dose per pulse effect (\( \alpha \)) is a polynomial function of the measured signal (\( N_i \)) to signal at 2.78 \( \times 10^{-4} \) Gy/pulse (\( N_o \)) ratio,

\[
\alpha = f\left( \frac{N_i}{N_o} \right)
\]

The corrected signal would be

\[
N_c = N_i \times \left( \frac{1}{\alpha} \right)^{N_i/N_o}
\]

The dose per pulse correction would be subsequently applied to the depth dose curve, penumbra, and IMRT dose measurements.

III.B. Percent depth dose

Figure 7 shows the measured percent depth dose profile using the strip detector compared with a Farmer ion chamber (NE-2571) measurement using solid water for a 10 × 10 cm² field size at source to surface distance of 100 cm. The photon energy was 6 MV. The CMRP DMG was corrected for dose per pulse effect using the method described in Sec. III A. The resulting depth dose curve agrees with the Farmer ion chamber measurements within 0.8% up to 20 cm depth in solid water. This difference is within the measurement uncertainty of the depth dose measurement. The \( D_{20/10} \) for the strip detector and the Farmer ion chamber were 0.573 and 0.569, respectively.

III.C. Stem effect

When the CMRP DMG was irradiated with the long axis of the field parallel to the cables, the increased in detector measurement was found to be 0.2% ± 0.1%. This shows that the stem effect is quite negligible for clinically useful field size.

III.D. Linearity

The strip detector response is linear with dose range of 3.89–311.05 cGy. The dose linearity verification was carried out at this range because it was deemed to be within the range of a normal IMRT dose per fraction. This is by no means the dynamic range limit of the detector as the acquisition rate of the detector readout system can be varied to accommodate large counts. For our measurement configuration, the slope of the best linear fit is 1597 counts/cGy with a regression coefficient of 1.00 (Fig. 8).

III.E. Uniformity

Figure 9 shows the detector response of each of the detector channels before and after uniformity correction is nor-
malized to 1 at the average of the channels. The results presented were the mean of three measurements and error bars represent the 95% confidence interval (CI) of the mean. In our case, the 95% CI was derived by taking the mean plus and minus 95% confidence coefficient (for two degrees of freedom) multiplied by the standard error of the mean. The results presented were the mean of three measurements and error bars represent the 95% confidence interval (CI) of the mean. In our case, the 95% CI was derived by taking the mean plus and minus 95% confidence coefficient (for two degrees of freedom) multiplied by the standard error of the mean.42

### III.F. Energy response

The energy response of the strip detector was studied using an orthovoltage unit and a linear accelerator. Figure 10 presents the energy response curve for the range of 50 kV–10 MV nominal energies (corresponding to 26.8 keV–2.97 MeV equivalent photon energies), normalized to 1 at 6 MV nominal photon energy. Error bars represent the combined uncertainties in the 95% CI of the mean for the measurements and the 0.5% calibration uncertainty. The filters for different nominal energies are made up of a combination of copper and aluminum filters. The photon equivalent energies for the orthovoltage energies were derived from the half value layers of the orthovoltage beams.43 For the 6 and 10 MV photon beams, the mean energies were taken from the Monte Carlo results of Mohan et al.44 The photon energy response curve showed an enhanced response at low energies up to six times the response at 6 MV. The detector showed an over-response at lower photon energies with the maximum dose response at 75 kV nominal photon energy. This is due to the increased photoelectric effect cross section in silicon at low energies.45 The energy dependence of the detector in free air geometry was expected as silicon is not tissue equivalent, and therefore is sensitive to changes in the energy spectrum. This is consistent with the literature.27,46,47 The effect of the energy dependence of this silicon detector in megavoltage beams will not be as significant as for measurements under the orthovoltage beams.27

### III.G. Angular response dependency

The angular response dependency of the CMRP DMG for the incident beam angle of 0°–90° was studied. There is an increased under-response measured by the strip detector compared to the dose measured by the CC13 ion chamber as the incident beam angle changes from 0° to 90°, as shown in Fig. 11. The angular response correction factor was derived by normalizing the measured dose of DMG to the dose measured by the CC13 ion chamber. Figure 11 shows the mean angular response of three points of measurement with a spatial distance of 1 cm from each other. The DMG readings were taken as an average of the channels located within ±1.5 mm of the measurement point of 2.8, 12.8, and...
22.8 mm along the length of the detector. The angular response at the point of ±1 cm away from the center is within 1% of the center. Therefore, it is deemed justified to apply the center angular correction factor onto all 128 detectors (strips) in the IMRT measurements. This is used as a first attempt to correct for the angular response dependence.

Angular response for the center channels of the strip detector is within 3.1% ± 0.1% for angle of 0°–45°. The largest angular response (28.1% ± 0.1%) was found to be at the gantry angle of 90° where the beam is parallel to the detector. The uncertainty reported was the standard error of the mean for the channels located within ±1.5 mm of the measurement point and the ion chamber measurements. For this particular setup, the measurement uncertainty for the measurement was <0.5% (standard error). The observed angular dependence may arise from the combined effects of the linear air cavity containing it within the phantom, the artifact created by ceramic substrate mounting, and the anisotropies inherent in the silicon chip. We will investigate the minimization of the angular dependence in future prototypes by eliminating the small air volume surrounding the silicon strip detector as well as mounting of the strip detector on the Kapton substrate, eliminating the attenuation effect of the ceramic substrate.

III.H. Penumbra measurement

The 80%–20% penumbra width of a 10 × 10 cm² field size for the 6 MV beam at 1.5 and 10 cm depths was measured using the strip detector and a comparison was made against measurements from Gafchromic EBT films (Fig. 12). The EBT film measurements were performed in a water tank for X-jaw penumbra measurements, while the MLC rounded leaf end penumbra was measured in solid water phantom. The strip detector and EBT film measured the penumbra at 1.5 cm depth for the secondary jaw to be 2.77 ± 0.04 and 2.71 ± 0.13 mm, respectively.

Cheung et al. measured a 80%–20% penumbra width of 3.0 ± 0.2 mm at 1.5 cm depth using the EBT film. Compared to their measurements, the strip detector seems to be able to resolve a narrower penumbra width. For the MLC rounded leaf end, the strip detector measures a 3.52 ± 0.04 mm penumbra width compared to the 4.6 ± 0.3 mm MLC penumbra width measured using EBT films by Butson et al. Penumbra width measured at 10.0 cm depth are 3.94 ± 0.04 mm for X-jaw and 5.60 ± 0.04 mm for the rounded left end of the MLC. This is again smaller than the measurement with the EBT films for the X-jaw of 4.50 ± 0.15 mm. Kron et al. investigated the 80%–20% penumbra width using thermoluminescent dosimetry extrapolating to infinitesimal small detector and found the penumbra width to be 3.2 mm at 1.5 cm depth and 4.2 mm at 10.0 cm depth. The results are summarized in Table I.

III.I. Clinical application in IMRT fields

The prostate IMRT plan was delivered with the Varian Clinac 21EX linear accelerator 6 MV photon beam. The beam angles used were 0°, 45°, 90°, 270°, and 315°; delivering a total of 596 MUs. The isocenter of the prostate IMRT plan was purposely shifted 4 cm to one side so that the CMRP DMG was positioned at an area of steep gradient. Figure 13 shows the DMG measured dose profiles obtained from individual beam angle and the total summation of all

<table>
<thead>
<tr>
<th></th>
<th>CMRP DMG (mm)</th>
<th>EBT film (mm)</th>
<th>Others (mm)</th>
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</thead>
<tbody>
<tr>
<td>X-jaw (symmetric) at 1.5 cm depth</td>
<td>2.77 ± 0.04</td>
<td>2.71 ± 0.13</td>
<td>3.0 ± 0.2</td>
</tr>
<tr>
<td>MLC (symmetric) at 1.5 cm depth</td>
<td>3.52 ± 0.04</td>
<td>4.08 ± 0.03</td>
<td>4.6 ± 0.3</td>
</tr>
<tr>
<td>X-jaw (symmetric) at 10.0 cm depth</td>
<td>3.94 ± 0.04</td>
<td>4.50 ± 0.15</td>
<td>4.2 ± 0.3</td>
</tr>
<tr>
<td>MLC (symmetric) at 10.0 cm depth</td>
<td>5.60 ± 0.04</td>
<td>6.61 ± 0.16</td>
<td>...</td>
</tr>
</tbody>
</table>

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*Reference 48.

bReference 49.

cReference 5.
the beam angles. The measured dose was corrected for uniformity, dose per pulse effect, and angular response dependency.

Figure 14 shows the comparison of dose profiles from the CMRP DMG, Gafchromic EBT film, and Pinnacle predicted dose. The error bars represent the reproducibility uncertainty in the measurements at the level of 95% CI of the mean.

The IMRT plan delivered was a step and shoot IMRT plan. The IMRT segments in each of the fields are quite discernable in our measurement due to the time gap for the MLC to move to position between each segment. However, due to the small detector size (2.56 cm), we may miss out some segments if they do not coincide with the location of the detector. The high spatial resolution of the detector together with the high temporal resolution of the readout and data acquisition system allow us to measure the temporal variation in the dose modulation within a single IMRT segment. Figures 15(a)–15(d) show the unique capability of this detector system. Figure 15(a) shows a single step and shoot IMRT field (gantry angle of 315°). The IMRT modulation of each segment is clearly depicted. The time for the delivery of each segment varies but within each segment the beam profile is constant and reproducible. Figure 15(b) shows the 3D (dose, time, and location) dose profiles of a full IMRT delivery, plotted temporally. A total of 1900 samples was acquired at the pulse width resolution of 0.1 s. Figure 15(c) showed the buildup of the cumulative dose, i.e., the temporal dose growth of the same IMRT delivery.

Various published literature have stressed the importance of the temporal dose resolution of IMRT as well as its efficacy in tumor cell kill. Insofar as reported in literature, the postulation of the reduced effect of tumor cell kill due to protracted IMRT delivery time was drawn from theoretical calculation and in vitro cell studies. Fowler et al. suggested that IMRT delivery that takes longer than half an hour to deliver per fraction might have a reduced cell killing effect. To the best of our knowledge, there is as yet no clinical studies that demonstrate a reduced treatment response from protracted IMRT delivery times. In the light of the lack of understanding about the temporal effect of radiation dose delivery in the clinical setting, the CMRP DMG may prove to be a valuable tool in investigating dose rate variation in real time. Although the high temporal resolution of CMRP DMG may not address the radiobiology effect of long IMRT delivery time (sometimes up to 30 min), it could, however, shed some light on the dose pattern variation at the level of 0.1 s or lower, which would be a valuable tool in the dosimetry of volumetric modulation arc therapy (VMAT). Figure 15(d) shows the temporal dose rate patterns (measured for each 10 s) for channels 8 and 127 of the total IMRT delivery extracted from the measurements. From this picture, it is clearly seen that there exist time intervals with similar patterns for both channels 8 and 127 when the dose rate is almost zero. This corresponded to the time between field deliveries. On the other hand, dose rate patterns during the field delivery are quite different, for example, during the time intervals of 40–60 and 120–130 s. Beam off during gantry movement time has been omitted from the figure. All of these variations are within the distance of 2.38 cm between the two points of measurements. To our knowledge,
this is the first report in radiation therapy of a dosimeter, allowing direct dose measurements in a particular phantom location in real time with 0.2 mm spatial resolution and 0.1 s temporal resolution.

**IV. CONCLUSION**

The CMRP dose magnifying glass comprising of a linear array of 128 silicon localized dosimeters (strips) each of them with an active area $20 \times 5000 \, \mu m^2$ and pitch of 0.2 mm was characterized and tested in a clinical IMRT delivery. For a 390-fold variation in the linac dose per pulse, the detector response changed by 23% compared to the CC13 ion chamber measurement. A correction function was introduced to correct for this effect in the measurements. The percent depth dose profile for the 6 MV photon energy matches closely with the measurement with Farmer ion chamber (within 0.8%) up to 20 cm depth in solid water. The stem effect of the detector was found to be negligible. The linearity of the detector with dose is excellent for the dose range of 3–300 cGy typical for of one fraction of IMRT delivery. The original strip detector had a $\pm 2\%$ variation in sensitivity between 128 channels including DAQ electronics that suggests very high reproducibility of the strip detectors with the planar microelectronic technology. A uniformity correction map has been obtained using flat field calibration methods leading to excellent uniformity of the response within 0.2%. The strip detector showed the typical energy dependency intrinsic for silicon at lower photon energies in free air geometry. Penumbra measurements were performed at the depths of $d_{max}$ of 1.5 and 10.0 cm for the 6 MV photon energy. The measurements were compared to the measurements of Gafchromic EBT film as well as other data published in literature for the same fields. The CMRP DMG with its physical spatial resolution of 0.2 mm derived a penumbra 80%–20% width of 2.77 mm at 1.5 cm depth and 3.94 mm at 10.0 cm depth, similar to the EBT film measurements as...
well as those measured by other dosimeters reported elsewhere. We demonstrated the application of the CMRP DMG in phantom dosimetry QA in a clinical IMRT delivery. The dose profiles from the measurement were compared to the planned dose from the treatment planning system as well as measurements made with Gafchromic EBT film at the same depth. The dose profile patterns measured by CMRP DMG were able to reproduce the dose profiles by the treatment planning system and film measurements. The average differences were 1.1% ± 1.8% (1 s.d.) and 1.0% ± 1.6% (1 s.d.) between measurements by the CMRP DMG with the Pinnacle predicted dose and the EBT film, respectively. The dose profiles agree fairly well with each other within the uncertainty in the measurements as well as the positioning uncertainty of 1 mm². The angular response of the strip detector is within 3.1% ± 0.1% for the angle of 0°–45°, while for the angle of 90° the response was 28.1% ± 0.1%. This is associated with the ceramic packaging of the detector as well as the inherent anisotropies in Si. The new generation of CMRP DMG placed on a Kapton pigtail is expected to have a minimal angular dependence and will be investigated soon. In addition to high spatial resolution of 0.2 mm, CMRP DMG is capable of providing integral dose profiles for each delivered segment or dose rate measurements at any channel with temporal resolution of 0.1 s. Such measurements for a typical prostate treatment delivery demonstrated a temporal pattern in dose rate delivery which is quite different within a short distance between points of measurements in the linear detector array. The combination of such properties in a dosimetry system for radiotherapy QA and for IMRT, in particular, is first reported in this paper.

The present design of the CMRP DMG is suitable for beam characterization and measurement of steep gradient dose profiles. The second generation CMRP DMG is placed on the Kapton pigtail. This design will allow for the 128-element detector to be stepped across horizontally or vertically to obtain a full dose profile magnification at points of interest. The 0.2 mm detector spatial resolution of the CMRP DMG would be suitable as a dosimetric tool for stereotactic radiotherapy QA as well. Application of this technology with the same spatial resolution for large radiation fields is limited by the number of readout channels and the detector-to-readout system connection logistic. However, within these limitations, a two dimensional prototype with a 16 × 16 microdiodes matrix and 2 mm pitch detector is currently under development. The final product will have four of these matrices (1024 diodes) arranged side by side within an active area of 64 × 64 mm².

For the current CMRP DMG, no special efforts were made for optimization of the dose rate response. Future p-Si based DMG detectors will be made dose rate independent by increasing initial p-Si concentration and preirradiation. This is to reduce the minority charged carrier injection level under higher dose rate irradiation, hence reducing the dose rate effect.

In conclusion, the CMRP silicon strip detector dose magnifying glass based on TERA ASIC DAQ system, with its high spatial resolution and the accompanying high temporal resolution readout system, is a novel tool for QA in IMRT dose delivery and VMAT dose delivery.

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