Characterization of a novel two dimensional diode array the “magic plate” as a radiation detector for radiation therapy treatment

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Purpose: Intensity modulated radiation therapy (IMRT) utilizes the technology of multileaf collimators to deliver highly modulated and complex radiation treatment. Dosimetric verification of the IMRT treatment requires the verification of the delivered dose distribution. Two dimensional ion chamber or diode arrays are gaining popularity as a dosimeter of choice due to their real time feedback compared to film dosimetry. This paper describes the characterization of a novel 2D diode array, which has been named the “magic plate” (MP). It was designed to function as a 2D transmission detector as well as a planar detector for dose distribution measurements in a solid water phantom for the dosimetric verification of IMRT treatment delivery.

Methods: The prototype MP is an 11 × 11 detector array based on thin (50 µm) epitaxial diode technology mounted on a 0.6 mm thick Kapton substrate using a proprietary “drop-in” technology developed by the Centre for Medical Radiation Physics, University of Wollongong. A full characterization of the detector was performed, including radiation damage study, dose per pulse effect, percent depth dose comparison with CC13 ion chamber and build up characteristics with a parallel plane ion chamber measurements, dose linearity, energy response and angular response.

Results: Postirradiated magic plate diodes showed a reproducibility of 2.1%. The MP dose per pulse response decreased at higher dose rates while at lower dose rates the MP appears to be dose rate independent. The depth dose measurement of the MP agrees with ion chamber depth dose measurements to within 0.7% while dose linearity was excellent. MP showed angular response dependency due to the anisotropy of the silicon diode with the maximum variation in angular response of 10.8% at gantry angle 180°. Angular dependence was within 3.5% for the gantry angles ± 75°. The field size dependence of the MP at isocenter agrees with ion chamber measurement to within 1.1%. In the beam perturbation study, the surface dose increased by 12.1% for a 30 × 30 cm² field size at the source to detector distance (SDD) of 80 cm whilst the transmission for the MP was 99%.

Conclusions: The radiation response of the magic plate was successfully characterized. The array of epitaxial silicon based detectors with “drop-in” packaging showed properties suitable to be used as a simplified multipurpose and nonperturbing 2D radiation detector for radiation therapy dosimetric verification. © 2012 American Association of Physicists in Medicine. [http://dx.doi.org/10.1118/1.3700234]

Key words: 2D diode array, magic plate, IMRT, transmission detector

I. INTRODUCTION

Radiation therapy has long been recognized as an effective method for treating cancer. In recent years, new treatment techniques such as the intensity modulated radiation therapy (IMRT) and more recently the volumetric modulated arc therapy (VMAT) have enable the escalation of dose with fewer side effects to the surrounding organs-at-risks.¹ ² The complexity of these new treatment techniques results in increased reliance on computers and machines to deliver the
treatment. The role of quality assurance and treatment verification becomes ever more important in the daily treatment delivery workflow. Traditionally, the dosimetric verification of an IMRT treatment involves measuring a point dose and/or a 2D dose distribution for comparison with the treatment plan predicted dose. This is usually achieved by means of point dose measurements in a low dose gradient region using an ion chamber and spatial fluence or dose profile measurement using two dimensional array dosimeters such as films, electronic portal imaging device (EPID) or 2D electronic arrays.

The use of 2D electronic arrays is gaining widespread use in the clinics. They provide efficient means of measuring dose at multiple locations in the field, real time feed back, and the ability to perform fluence/planar dose comparison simultaneously. Examples of some of the commercially available 2D detector arrays are the ionization chamber based I’mRT MatriXX, COMPASS 2D transmission detector (IBA, Germany) and 2D-ARRAY Type 10024 (PTW Freiburg, Germany) and semiconductor based MAPCHECK (Sun Nuclear, Melbourne, Fl). Efforts are also being made to utilize existing electronic portal imaging devices (EPID) as 2D detector arrays.

In the framework of the European Integrated Project MAESTRO, a dosimeter utilizing pixelated silicon detectors on epitaxial silicon was designed and prototyped. The preliminary characterization of this device was reported by Talamonti et al. and Bruzzi et al. A single epitaxial diode is an attractive alternative to a conventional silicon detector due to the thin epitaxial layer which is deemed to be more radiation hard. This technique has the ability to reduce the detector sensitive thickness while producing well defined sensitive volume. As a result, the detector is independent of radiation dose without compromising mechanical rigidity of the detector for mounting. This is a clear advantage in radiation therapy.

The use of epitaxial technology for radiation detectors used in medical radiation therapy is still rare due to the difficulty of the fabrication technique. A novel 2D diode array based on the epitaxial technology and the “drop-in” mounting technique was developed at the Centre for Medical Radiation Physics (CMRP), University of Wollongong. It was designed as a dosimetric tool for Intensity modulated radiation therapy (IMRT). The “drop-in” mounting of the epitaxial diode attempts to minimize energy dependence and improve the angular response of the diodes. This was achieved by using thin aluminum contacts on the periphery of the 0.5 × 0.5 mm² n⁺ ion implanted regions using a special technology. The lead of the diode is fully embedded in the Kapton substrate. The diode’s face is aligned with the Kapton substrate’s front surface. This technology avoids the use of high Z metal contacts (wire bonding) above the sensitive area and copper contact plate below the die.

The MP was designed to be used as a transmission detector and as a 2D planar detector for dose measurement in solid water phantoms. The MP packaging needs to be modified to allow each of these applications.

When used in transmission mode, the MP was sandwiched between two pieces of clear plastic sheets (100 μm, solid water plates (1 mm) and black plastic sheets (80 μm). The clear plastic sheets protect the MP diodes from dust and accidental contacts. The two solid water plates serves three purposes, (i) to shield the detector from ambient lights, (ii) for mechanical strength and protection of the MP, and (iii) as scattering material to increase the signal generated in the diodes. Finally, a black plastic sheet was used to wrap the MP and the solid water plates together to reduce light leaking to the detector [Fig. 2(a)]. The whole assembly was then clamped between two 6 mm Perspex slabs. This Perspex frame is then screwed onto a modified total body irradiation (TBI) tray which can be attached onto the linac accessory slot. The source to detector distance (SDD) in the accessory slot is 58 cm on a Varian 2100iX linac.

When used as a 2D planar diode array for dose measurement in a phantom, the MP was sandwiched between two pieces of 5 mm solid water plates which were machined to fit the Kapton substrate. The width of the solid water plates was 160 mm. This allows the MP to be used in conjunction with the I’mRT phantom (IBA Dosimetry) [Fig. 2(b)].
Prior to the use of the device in a clinical setting, the MP was characterized using similar tests to those carried out when testing the suitability of detectors as dosimeters in the radiation therapy environment. The parameters studied were the radiation damage effect, dose per pulse dependence, percent depth dose, dose linearity, energy dependence, angular dependence, field size dependence, and beam perturbation effect. The detector characterization was done onto the whole MP array. However, since the results were benchmarked against conventional single volume dosimeters such as ionization chambers and diodes, the response of the central MP diode was often used as the selected diode. This is done for most of the studies except for the angular dependence study where the individual diodes were also evaluated for their response when irradiated from various beam angles. The suitability of the use of a single MP diode for comparison was evaluated as part of the radiation damage studies.

II.A. Radiation damage studies

The sensitivity of silicon radiation diodes can change over time. This is because ionizing radiation creates defects in the silicon lattices. Depending on the type of radiation source, either point defects (in case of gamma radiation) or cluster defects (such as in a fast neutron field) were created. These defects become recombination centers for the minority charge carriers, reducing the minority carriers’ lifetime and the sensitivity of the detector. This is because the diffusion length is reduced, hence the sensitive volume of the diode is reduced. The effect of radiation damage on silicon diodes are also manifested in the dose per pulse dependence, increased dark current and temperature dependence. The effect of radiation damage on silicon diodes has been shown to decrease the diode sensitivity.

The MP was given a 41.5 kGy irradiation prior to the characterization exercise. The dose was delivered in two parts. The first 1.3 kGy was delivered using 6 MV photons from a medical linear accelerator and the rest was delivered by a high dose rate (3 kGy/h) cobalt-60 source at the Gamma Technology Research Irradiator (GATRI) facility, Australian Nuclear Science and Technology Organisation (ANSTO).

II.B. Dose per pulse dependence studies

Dose per pulse (dpp) dependency refers to the change of detector’s sensitivity due to the change of dose rate in a pulsed radiation beam such as those delivered by a medical linear accelerator. The dose rate during each pulse is sufficiently high to cause a variation in silicon detector response. It is therefore important to characterize new detectors for dose per pulse sensitivity. This effect was first reported by Rikner and Grusell. They found that n-type silicon diodes are more sensitive to dose per pulse variation compared to p-type diodes, showing an increase in sensitivity with increased dose per pulse. They also found that preirradiation of the diode also reduced the dose per pulse dependence.

The dose per pulse dependence measurements were made for the range of $1.71 \times 10^{-6}$ to $3.79 \times 10^{-4}$ Gy/pulse. This was achieved with five measurement setups involving the use of attenuators such as lead (Pb) blocks and irradiation under a closed multileaf collimator (MLC). Table I lists the measurement setups used, including a pseudo dose per pulse measurement made in free air geometry. The MP was irradiated with a $10 \times 10$ cm$^2$ field size beam for a fixed number of monitor units (MUs).

In three of the setups, two Pb blocks with thicknesses of 2.0 and 4.0 were placed in the linac beam supported by a block tray. The linac beam was completely blocked by the Pb. However, the use of Pb attenuators and closing the MLC hardens the linac beam spectrum. This may result in a
The diode sensitivity, $S$, is the ratio of the charge measured by the silicon diode to the charge measured by the ion chamber at the same position and setup. The dose per pulse response was taken as the detector sensitivity of the MP or the EFD-3G diode at any dose per pulse normalized to 1 at the dose per pulse of $2.78 \times 10^{-4}$ Gy/pulse.

II.C. Percent depth dose and dose linearity measurements

The percentage depth dose of a 6 MV photon beam was measured in a solid water phantom. The MP is sandwiched between slabs of solid water phantoms of variable thickness. A $10 \times 10$ cm$^2$ field size was used and measurements were made at selected depths from 1.2 to 300 mm with the source to surface distance of 100 cm. The minimum build up of 1.2 mm comprised of the 1 mm solid water, 0.1 mm clear plastic sheet, and 0.08 mm black protective sheet shown in Fig. 2. Due to the thin epitaxial layer ($50 \mu$m), the MP can be used to measure the buildup region of the depth dose curve. The result was compared with parallel plane ion chamber measurements for the buildup region (0–15 mm depth) and the standard data measured using a CC13 ion chamber (3–300 mm depth) in water.

For the dose linearity experiments, the MP was set up in a solid water phantom at the SSD of 100 cm and depth of 1.5 cm in a $10 \times 10$ cm$^2$ radiation field. The dose linearity was measured for the dose range of 5–1000 cGy.

II.D. Energy dependence measurement

The energy response of the MP was studied using an orthovoltage machine (Gulmay DX 3300) for the nominal

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**Table I.** Measurement setup for the dose per pulse measurement.

<table>
<thead>
<tr>
<th>Measurement setup</th>
<th>Setup description</th>
<th>SDD range (cm)</th>
<th>Dose per pulse range (Gy/pulse)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Free air geometry</td>
<td>MP sandwiched between two pieces of 1 mm solid water phantom and suspended in air</td>
<td>60.5–100.0</td>
<td>…</td>
</tr>
<tr>
<td>Open field</td>
<td>1.5 cm depth in solid water phantom with a $10 \times 10$ cm$^2$ open field</td>
<td>87.0–154.4</td>
<td>$1.20 \times 10^{-4} – 3.79 \times 10^{-4}$</td>
</tr>
<tr>
<td>2.0 cm Pb attenuation</td>
<td>1.5 cm depth in solid water phantom, radiation beam blocked by 2.0 cm Pb block</td>
<td>87.0–154.4</td>
<td>$3.53 \times 10^{-5} – 1.17 \times 10^{-4}$</td>
</tr>
<tr>
<td>4.0 cm Pb attenuation</td>
<td>1.5 cm depth in solid water phantom, radiation beam blocked by 4.0 cm Pb block</td>
<td>87.0–154.4</td>
<td>$1.33 \times 10^{-5} – 4.45 \times 10^{-5}$</td>
</tr>
<tr>
<td>6.0 cm Pb attenuation</td>
<td>1.5 cm depth in solid water phantom, radiation beam blocked by 6.0 cm Pb block</td>
<td>87.0–154.4</td>
<td>$4.80 \times 10^{-6} – 1.70 \times 10^{-5}$</td>
</tr>
<tr>
<td>Under closed multileaf collimator (MLC)</td>
<td>1.5 cm depth in solid water phantom, MLC closed</td>
<td>87.0–154.4</td>
<td>$1.71 \times 10^{-6} – 5.81 \times 10^{-6}$</td>
</tr>
</tbody>
</table>

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**Table II.** Specifications of the dosimeters used.

<table>
<thead>
<tr>
<th>Dosimeter</th>
<th>NE-2571</th>
<th>CC13</th>
<th>EFD-3G</th>
<th>MP</th>
</tr>
</thead>
<tbody>
<tr>
<td>Detector type</td>
<td>Farmer type ion chamber</td>
<td>Compact chambers</td>
<td>Commercial diode</td>
<td>MP</td>
</tr>
<tr>
<td>Device details</td>
<td>—</td>
<td>—</td>
<td>Low resistivity p-type silicon diode</td>
<td>High resistivity p-type epitaxial diode</td>
</tr>
<tr>
<td>Sensitive volume</td>
<td>$0.6 \text{ cm}^3$</td>
<td>$0.13 \text{ cm}^3$</td>
<td>$2.9 \times 10^{-4} \text{ cm}^3$</td>
<td>$1.25 \times 10^{-3} \text{ cm}^3$</td>
</tr>
<tr>
<td>Sensitive area</td>
<td>$6.3 \text{ mm} \ominus$</td>
<td>$6.0 \text{ mm} \ominus$</td>
<td>$2.5 \text{ mm} \ominus$</td>
<td>$0.5 \times 0.5 \text{ mm}^2$</td>
</tr>
<tr>
<td>Sensitive thickness/length</td>
<td>$24.0 \text{ mm}$</td>
<td>$5.8 \text{ mm}$</td>
<td>$0.06 \text{ mm}$</td>
<td>$0.05 \text{ mm}$</td>
</tr>
<tr>
<td>Physical size</td>
<td>—</td>
<td>—</td>
<td>$7 \text{ mm} \ominus$</td>
<td>$1.5 \times 1.5 \times 0.05 \text{ mm}^3$</td>
</tr>
</tbody>
</table>
energies of 75–250 kV and a Varian Clinac 21EX linear accelerator (Varian, Palo Alto, USA) for the photon energies of 6 and 10 MV. For the 50–150 kV tube voltages, 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 voltages, 100 mm diameter circular applicator was used with detector positioned at the focus to surface distance of 512 mm. For the high energy photons, the detector was irradiated at the depth of maximum.

Energy response of the MP was evaluated for four different geometries (Table III) and the readings were normalized to 1 at the energy of 6 MV. The “face-up” and “face-down” configurations are shown in Fig. 3.

II.E. Angular dependence measurement

Most of the large area 2D detector arrays were originally designed for orthogonal incidence beam. Using these arrays, IMRT treatment verifications are mostly performed with all beams delivered normally incident on the array from a fixed gantry angle. However, recently, various groups have explored the use of these arrays for rotational beam measurements. The use of 2D array detectors for rotational beam measurements required the characterization of the angular dependence of the devices. Bhadwaj et al. showed that the MatriXX system (IBA Dosimetry, Germany) has a maximum angular response of 7.7% when beams were delivered at gantry angle 180°. Jursinic et al. found that the metal contacts of the diode resulted in ±20% of angular dependence in the MAPCHECK (Sun Nuclear, Melbourne, Fl) detector. Modification of the detector design by inserting copper pieces to offset the diode asymmetry and Lucite sheets to fill in air gaps reduced the angular dependence to ±2%. The angular response of the MP was carried out using an IMRT phantom (IBA Dosimetry, Germany). The MP was sandwiched between two pieces of 5 mm thick solid water phantoms. The MP was positioned at the depth of 9 cm in the phantom at the SDD of 100 cm. A fixed number of MUs were delivered with a 16 × 16 cm² radiation field size. The larger field size was chosen in order to cover all the 11 × 11 MP diodes extending over an area of 10 × 10 cm². A Varian Clinac iX (Varian, Palo Alto, USA) linear accelerator was used for this measurement. The angular dependence of the MP was studied with the MP in the face-up configuration for the gantry angles of 0° to 180° at 15° intervals. The asymmetric design of the epitaxial diode may result in directional dependence when the beam is entering from the face-up or face-down configuration. In addition, for the gantry angles 120°–180°, the radiation beam passing through the linac couch may be attenuated. This may result in a different silicon response due to energy dependence of the silicon. To isolate the angular response of the epitaxial diode due to anisotropy from energy dependence, the MP was irradiate with the face-down configuration for the gantry angles 0°–90° and compared with the corresponding angular response made with the MP in the face-up configuration.

<table>
<thead>
<tr>
<th>Geometry</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>A face-up, in phantom&lt;sup&gt;a&lt;/sup&gt;</td>
<td>MP sandwiched between two pieces of 1 mm solid water plates and position in the face-up orientation on top of 11.6 cm of solid water backscatter material</td>
</tr>
<tr>
<td>B face-up, free air geometry</td>
<td>MP sandwiched between two pieces of 1 mm solid water plates in the face-up orientation, without the backscatter material</td>
</tr>
<tr>
<td>C face-down, in phantom&lt;sup&gt;a&lt;/sup&gt;</td>
<td>MP sandwiched between two pieces of 1 mm solid water plates and position in the face-down orientation on top of 11.6 cm of solid water backscatter material</td>
</tr>
<tr>
<td>D face-down, free air geometry</td>
<td>MP sandwiched between two pieces of 1 mm solid water plates in the face-down orientation, without the backscatter material</td>
</tr>
</tbody>
</table>

<sup>a</sup>Build up thickness is photon energy specific. For the 6 and 10 MV photon energies, the measurement depths were at d<sub>max</sub>.

![Fig. 3. Schematic drawing of the face-up and face-down configurations for MP diode.](image-url)
Gafchromic EBT2 film (ISP, Wayne, NJ) was used as the benchmarking dosimeter in this study. The EBT2 films were cut into sections of 101.5 × 120.0 mm² and positioned at the center of the IMRT phantom at the depth of 9 cm. The EBT2 films were scanned with an A3 flatbed scanner (Epson Expression 10000XL). Each film was scanned six times. Only the last three scans were kept for image analysis. This was to ensure that the scanner was sufficiently warmed up and thus ensuring consistency in the scanned films. The films were scanned in 48-bit RGB color with the scanning resolution of 96 dpi (equivalent to pixel size of 0.265 mm). Care was taken to scan the film in the same orientation at the center region of the scanner to reduce scanner induced non-uniformity. A set of calibration films was also irradiated in the center of the IMRT phantom at the depth of 9 cm. The SDD of 101.5 and 1.5 cm depth in a solid water phantom.

The angular response of the MP diodes was evaluated solely in the azimuth direction. The angular response of each diode was evaluated. However, for simplicity of presentation in this paper, the angular coefficients for the 11 diodes positioned along the same column were averaged. Both data sets were fitted with 2nd degree polynomial fits.

The angular coefficient of the MP diode was taken as the ratio of the signal measured by the MP over the EBT2 dose values [Eq. (1)].

\[
\text{Angular coefficient, } k_{ai} = \frac{\text{MP}_i}{\text{EBT2}_i}, \quad \text{(units = counts/cGy)},
\]

where, the index \(i\) represent the \(i\)th detector column.

The angular response of the MP diode was taken as the ratio of the angular coefficient of an arbitrary gantry angle, \(\theta\) to the angular coefficient of the gantry angle \(0^\circ\) [Eq. (2)].

\[
\text{Angular response, } C_{ai} = \frac{k_{a_{i0}}}{k_{a_{i}}},
\]

II.F. Field size dependence measurement

The field size dependence of the MP was evaluated for selected square field sizes from 5 × 5 to 40 × 40 cm². Measurements were made in two conditions; (a) in a solid water phantom at the SDD of 101.5 cm, 1.5 cm depth, and (b) in free air geometry at the SDD of 58 cm. The SDD of 58 cm represents the position of the MP when used as a transmission detector, attached to the linac’s accessory slot. Only a single diode located at the central axis was used to evaluate the field size dependence. MP measurements were compared to the standard data measured using a Farmer ion chamber at the SDD of 101.5 and 1.5 cm depth in a solid water phantom.

II.G. Beam perturbation measurement

The use of the MP as a transmission detector may attenuate the beam or influence the secondary electron contamination of the photon beam. This necessitates the investigation of the radiation beam perturbation effect. Currently, there are three other transmission detectors that were reported in the literature; the DAVID system, the Integral Quality Monitoring system (IQM), and the IBA COMPASS detector. All of these devices showed increased surface dose and dose within the build up region. For the dose beyond the \(d_{max}\), the presence of transmission devices caused beam attenuation up to 7%. The beam perturbation study for the MP was carried out to study two possible influences of the MP on the radiation beam; (a) the surface dose and dose within the build up region of a depth dose curve, and (b) the transmission factor at the depth of 10 cm in a solid water phantom.

II.G.1. Surface dose and build up region measurements

The surface dose and the build up region (0–15 mm depth) of a 6 MV photon beam were measured. Measurements were made in a solid water phantom using a Markus type (PTW, Freiburg, Germany) parallel plate ion chamber. Markus type chambers were known for their over-response due to the small guard ring. Various methods were developed to correct for this over-response. In this work, the Rawlinson correction was used to correct for the Markus chamber measurements. The Markus chamber used was an original Markus chamber, type 23343. The chamber dimension used for the Rawlinson correction calculation was obtained from Chen et al. Two field sizes (20 × 20 and 30 × 30 cm²) were measured at two sources to surface distances (SSDs) of 80 and 90 cm. Measurements were made with the MP mounted in the accessory tray (MP field) and with an open field. The measurements were normalized to the depth of maximum dose.

II.G.2. Transmission factor measurements

When a block tray or physical wedge was placed on the linac accessory tray, a tray/wedge transmission factor was applied to correct for the attenuation of the photon beam quality due to the tray/wedge. The tray transmission factor was defined as a ratio of the dose in a phantom with and without a tray in the beam for the same number of monitor units delivered at a fixed source to detector distance and depth in phantom. The transmission factor of the MP was measured at the SSD of 100 cm and the depth of 10 cm in solid water phantom for three field sizes (10 × 10, 20 × 20, and 30 × 30 cm²). A CC13 ion chamber was used for this measurement.

III. RESULTS

III.A. Radiation damage effect

Figure 4 shows the detector sensitivity as a function of accumulated radiation dose. The error bars represent the ±1...
standard deviation of the channels located within the central 20%–80% region of the 11 × 11 MP detector. The positive charge build up at the Si-SiO₂ layer due to the accumulated radiation caused the apparent increase in the sensitivity of the detector by a factor of two after a dose of 41.5 kGy. Figure 5 shows the medium term reproducibility of the central diode of the MP up to 64 days, postirradiation with 41.5 kGy. The variability of the individual detectors located within the central 20%–80% region of the MP was 2.1% (coefficient of variation). This demonstrates that the use of a single MP diode for detector response comparison is within reasonable uncertainty. This study also shows that the MP sensitivity was stable after it was given a dose of 41.5 kGy. No observation in radiation damage of the kapton was recorded.

III.B. Dose per pulse dependence

Figure 6 shows the dose per pulse response measurement for a highly doped commercial p-type Si diode (EFD-3G) and the epitaxial diode of the MP, normalized to the dose per pulse of 2.78 × 10⁻³ Gy/pulse (estimated dose rate at the depth of 1.5 cm and SSD of 100 cm). For measurements with closed MLC, the large uncertainty (± 15.5%) is due to the positioning error leading to MLC leakage. The use of attenuators such as Pb pieces and 10 cm thick MLC causes a change in the beam spectrum. This is clearly demonstrated in the discrepancies between the data points measured by the last measurement point under 6 cm thick Pb piece and the first measurement point under closed MLC. Figures 7(a)–7(f) shows the same dose per pulse response measurements as a function of source to detector distances (x-axis) for six measurement setups. The EFD-3G is an unshielded diode with a thin sensitive volume of 0.06 mm. The diode showed dose per pulse independence for almost all dose rates. However, the measured responses in free air geometry showed a dropped in diode sensitivity relative to the ion chamber measurements.

The epitaxial diode exhibited a larger dose per pulse dependence compared to the EFD-3G diode. The onset of the drop in the epitaxial diodes’ sensitivity was observed at the dose per pulse of 1 × 10⁻⁴ Gy/pulse.

The MP’s dose per pulse dependence was shown as a drop in the diode sensitivity relative to the ion chamber measurements at the same dose rate. The same trend was also observed in the EFD-3G diode in the free air geometry setup. However, this trend was in contrast with the dose per
pulse dependence reported in most literature, which showed increased diode sensitivity as the dose rate increased. The reasons for both of these trends (either increased or decreased of diode sensitivity with increased dose rates) have been detailed in various literature.51–55 It will be qualitatively discussed in this paper.

The dose rate dependence of commercial silicon diodes to pulsed ionizing radiation was widely reported.28,29,31,52,56 Most literature reported an increase in diode sensitivity with increased dose rates. The explanation for this effect was based on the Shockley-Read-Hall (SRH) recombination model.

Figures 7(a)–7(c) shows the normalized ratio of the measurements made using the lead pieces. Both the MP and the EFD-3G diode showed dose rate independence. Figure 7(d) shows the normalized ratio of the measurements made in phantom with a 10×10 cm² open field. The EFD-3G appeared to be dose rate independent while the epitaxial diodes showed a decrease in sensitivity at higher dose rates. The dose rate response of the epitaxial diode seen here may be rare in the existing publication on silicon radiation diode used for medical radiation dosimetry. Nonetheless, this behavior is not unique to the epitaxial diode as it was reproduced by the commercial EFD-3G diode in the measurements made in free air geometry, as observed in Fig. 7(e). This phenomenon was explained by Alexander53 and Fjeldy et al.54

At extreme high dose rates (excess minority carriers >> the density of the majority charge carriers), the SRH recombination model would start to fail. This is because Auger recombination becomes more dominant.53,54 The onset of the Auger recombination is dependent on the doping concentration with higher resistivity material beginning at a lower dose rate than a lower resistivity material.53

Figure 7(e) shows the measurements made in free air geometry. Both MP epitaxial diode and the EFD-3G diode showed a decreasing sensitivity with increased dose rates, albeit at a much higher sensitivity (up to a factor of 1.4) relative to the normalization point (2.78×10⁻⁴ Gy/pulse). In reality, the injection level in the free air geometry setup is much higher than the in phantom open field setup due to the shorter SDD. However, since dose is defined in phantom at a certain depth and SSD, the actual dose in the free air geometry measurement cannot be determined due to three factors,

(i) Charge particle equilibrium (CPE) condition does not exist in the free air geometry. The measured dose in air by a large diameter ion chamber without a build up cap and dose in silicon would also be largely affected by the Si/air mass energy absorption coefficient ratio.

(ii) Without the CPE in a full scattering condition, the actual dose measured by the silicon diode would comprise of only a fraction of the dose at dmax. However, since the ion chambers and the diodes have different sensitive volumes, the effect of volume averaging in the larger detectors would give a lower reading.

(iii) Measurement of the fluence at short SDD also exposed the detectors to a larger portion of the low energy photons scattered by the components in the linac head.57 This would induced an over-response in the silicon detectors at photons energy <150 keV.

Figure 7(f) shows the S/S₀₂.₇₈×₁₀⁻⁴ ratio of the measurements made in phantom under a closed MLC. Both the MP and the commercial diode showed dose per pulse independence for the range of SDD measured. The MP appears to be under-responding by 8.9% while the commercial diode over-responded by 7%. The large error bar of 15.5% showed the measurement error due to the MLC leakage. This value was obtained by taking multiple measurements with the diode position translated perpendicular to the long axis of the MLC.
leaves by up to 0.3 mm, thereby measuring variation between the dose directly under a MLC leaf and the MLC leakage dose. The same error was applied onto MP measurements.

The error bars representing the uncertainties for all (except for MLC related measurements) MP and commercial diode measurements were 0.2% and 0.4%, respectively. This represents the maximum uncertainty of the measurements which comprised of the 1 standard deviation of the mean of three measurements for either MP (or diode) and the ion chamber added in quadrature. For measurements with closed MLC, the uncertainty due to the positioning error due to MLC leakage was included, giving a total uncertainty of 15.5%.

The need to apply a dose per pulse response correction to the measured data is determined by the measurement setup and the magnitude of the dose per pulse effect. When the MP is used as a transmission device at a fixed position downstream of the MLC, no dose per pulse correction is required for the measured fluence since there is no variation of the source to detector distance. When the MP is used as a 2D planar dose array in the I’mRT phantom at a depth of 9 cm with an isocentric technique, the SDD to the central diode remains the same SDD of 100 cm. In the worst case scenario, where the radiation beam is delivered from gantry angle 90° or 270°, the distance between the row of detector nearest to the linac source and the row furthest away from the linac source is ±5 cm at a depth of 9 cm. From a tissue-maximum-ratio (TMR) chart, this represents a change of ±14% of the dose, which correlates to ±1.5% of difference in dose per pulse correction. This is within the ±2% of the field flatness and symmetry. In this case, the dose per pulse correction may not be necessary.

In situations where the dose per pulse correction was required, the dose per pulse correction factor can be derived from these measurements. The method used for the dose per pulse correction was described in Wong et al.58,59

### III.C. Percent depth dose and dose linearity

Figure 8 shows the depth dose curve measured with the MP and the CC13 ion chamber. The MP appears to be slightly over-responding compared to the ion chamber measurements for all depths beyond the d_{max} with the maximum over-response of 1.74% at 10 cm depth. This may be due to the dose per pulse response of the MP. The under response of the diode at high dose rates resulted in an underestimation of the dose at d_{max} and overestimation of the dose beyond d_{max}. The depth dose curve was subsequently corrected for the dose rate effects and the resulting curve agrees with the CC13 data within 0.7%. The dose per pulse response of silicon diode affecting the percent depth dose measurements were also reported by Wilkins et al.28

Figure 9 shows the buildup region, comparing the MP measurement with the parallel plane ion chamber and the...
CC13 ion chamber. The MP data were dose per pulse corrected. Based on this figure, the water equivalent depth of the MP diodes at 1 mm depth was 1.9 mm.

The dose linearity was excellent for the measured dose range of 5–1000 cGy (goodness of fit, $r^2 = 1$).

### III.D. Energy dependence measurement

The energy response of the MP was studied using an orthovoltage unit and a linear accelerator. Figure 10 presents the energy response curve for the range of 75–10 MV nominal energies (corresponding to 26.8–2.97 MeV equivalent photon energies), normalized to 1 at 6 MV nominal photon energy (face-up, in phantom geometry). 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 filters.\(^6\) For 6 and 10 MV photon beam, the mean energies were taken from Mohan et al.\(^6\) Monte Carlo results.

The photon energy response curve showed an enhanced response at low energies up to 7.5 times of the response at 6 MV for the face-down, in phantom geometry. The detector showed and 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.\(^6\)

Energy dependence of the detector in the free air geometry was expected as silicon is not tissue equivalent and therefore sensitive to changes in the energy spectrum. MP used in the free air geometry had a lower signal response due to the lack of backscatter electron contribution.

The detector’s response in the face-up configuration was lower than in the face-down configuration. This may be due to the presence of the 0.375 mm silicon substrate. However, this silicon substrate affects the detector response in different ways depending whether the detector is in the free air geometry or used with the solid water phantom.

#### III.D.1. Detector irradiated faced up with the solid water phantom

For the low electron energies of $<150$ keV, the range of electrons in silicon is less than 0.35 mm. All back scattered electrons generated in the phantom were stopped in the silicon substrate. Hence, the detector response in the face-up configuration was lower.

![Fig. 8. Depth dose curve for a 10 × 10 cm$^2$ field size of a 6 MV photon energy measured with a CC13 ion chamber and the MP.](image1)

![Fig. 9. Build up region of the depth dose curve, comparing the MP, CC13 and parallel plane ion chamber measurements.](image2)
III.D.2. Detector irradiated faced down in the free air geometry

The increased of the detector response at with the face-down configuration may be due to the dose enhancement effect of the higher Z of silicon, i.e., additional number of electrons were forward scattered from the silicon substrate to the sensitive volume of the epitaxial layer downstream. With decreasing photon energies, this effect is gradually reduced as the competing effect of the x-ray attenuation due to the 0.375 mm silicon substrate.

Figure 11 showed the energy response in the megavoltage (MV) region. Comparing to the energy response at low photon energies, the energy response of the epitaxial diodes for measured in the solid water phantom was reversed. The face-down configuration showed a lower response compared to the face-up configuration for the measurements in the phantom. For the in phantom measurement at SSD 100 cm, the epitaxial diode in the face-down configuration underresponded by 9.7% ± 2.1% and 7.0% ± 2.1% for the 6 and 10 MV, respectively. This may be because the effect of the x-ray and secondary electrons attenuation due to the 0.375 mm silicon substrate dominates the detector response, whilst enhancement is the dominant effect in the free air geometry configuration.

In the free air geometry, the MP was sandwiched between two pieces of 1 mm solid water phantom. This setup does not have electronic equilibrium; hence, the signal measured by the MP comprised only of a fraction of the dose at a full scattering, CPE condition. In this case, the measured signal were 53.0% ± 1.5% and 37.7% ± 2.5% (mean error) for the 6 and 10 MV photon energies, respectively.

III.E. Angular dependence measurement

Figure 12 shows the mean angular response of the MP diodes located on three detector columns representing the center and two lateral edges of the MP (–5, 0, and +5 cm off axis distance). The angular response of the MP diodes was <2.7% for gantry angles 0°–60° (corresponding mirror angles of 300°–360°) and 4.5% at gantry angle 75° (mirror angle of 285°). Due to asymmetric geometry of the MP diode and the inherent silicon anisotropy, the mean angular response for gantry angles 90°–270° was 10.5% ± 0.7%.
At the gantry angle of 180°, the mean angular response was 10.8% ± 0.7%.

Response anisotropy is described as the nonsymmetrical geometry of the silicon diode. The MP epitaxial silicon diode has a physical size of 1.5 × 1.5 mm² and a total thickness of 425 μm, of which the epitaxial layer (the thickness of the sensitive volume) made up the top 50 μm. This resulted in a nonsymmetrical surrounding of the sensitive volume by a passive silicon volume where charge collection is absent. In addition, the attenuation of the secondary electrons or dose enhancement is also not the same for different beam directions.

The uncertainties of the measurements were taken as the ±1 standard deviation of the mean of 11 diodes and the equivalent pixel positions on the film. The average uncertainty was ±2.3% and ±0.7% for the film and MP measurements, respectively.

III.F. Field size dependence

Figure 13 shows the field size dependence of the MP at two different SDDs. The output factor measured using the Farmer chamber was also included for comparison. The field size dependence of the MP measured at SDD 101.5 cm agreed well with the Farmer chamber measurement within 1.1%. The field size dependence of the MP measured at the SDD of 58 cm showed a much steeper slope. This may be because at shorter SDD and in the free air geometry, the amount of collimator scatter and phantom scatter contribution to the field size factor was vastly different to those measured in phantom with full scattering condition.

In a phantom, the response of the silicon diode was determined by the secondary electrons whereas in the free air geometry, the diode response was determined by photons, i.e., the mass energy absorption coefficients of the silicon. The resulting energy dependence is due to scattered radiation. At smaller field sizes, the lack of phantom scatter resulted in a much lower field size factor, whereas for field size >10 × 10 cm², the higher contribution of the collimator scatter due to the shorter SDD resulted in a higher field size factor. Asuni et al. performed Monte Carlo studies on the contaminant electrons due to transmission detector. They found that the transmission detector increased the contaminant electrons at shorter SSD of 70 cm and increasing with larger field sizes. However, most contaminant electrons are of low energies with large angular spread. Hence, it does not contribute much to the dose at large SSD (e.g., 100 cm). They also looked at the contribution of contaminant electrons by various components in the linac head. For an open field, the phase space file showed that the major difference in the contributors were the secondary jaws and the air column between the linac and the surface of the phantom. At SSD of 70 cm, the jaws contributed up to 15% more contaminant electrons whereas the air column contribution was 16.8% higher at...
SSD of 100 cm for a 10 × 10 cm² field size. In addition, the mean energy scored for the short SSD was also found to be slightly lower.

### III.G. Beam perturbation studies

#### III.G.1. Surface dose and build up region of the depth dose curve

The presence of the MP as a transmission detector increases the surface dose and the dose within the buildup region of the depth dose curve (Fig. 14). The surface dose and build up region under the MP was compared with the open field measurements collected with a parallel plate ionization chamber. The increased surface dose and dose within the buildup region was attributed to the increased electron contamination due to the presence of the MP. Table IV shows the surface dose difference due to the presence of the MP in percentage. For the 20 × 20 cm² field size, the surface dose increased by 3.4% and 7.3% for the SSD of 90 and 80 cm, respectively. This was 4.7% and 10.9% lower than the COMPASS system¹¹ for the same field size and SSDs. The electron contamination increases with larger field size and with shorter SSDs. For the largest field size investigated at the SSD of 80 cm, the presence of the MP would increase the dose to the superficial tissues by 12.1%.

#### III.G.2. Transmission factor

The transmission factors for three field sizes of 10 × 10, 20 × 20, and 30 × 30 cm² were measured. Average transmission factor was 0.990 ± 0.002 (1 standard deviation) for 6 MV beam. Compared to the other transmission detectors, the DAVID and IQM systems attenuate a 6 MV beam by 7% while the COMPASS system reported a 3.3% beam attenuation. The low transmission factor found while using the MP in-line with the radiation beam shows the feasibility of this device as a transmission dosimeter.

### IV. CONCLUSIONS

The 2D detector, magic plate, comprises of 11 × 11 epitaxial silicon diodes with a sensitive volume of 0.5 × 0.05 mm³ and is mounted on a 0.6 mm thick Kapton substrate using the “drop-in” technology. The radiation response and basic characterization of the MP was carried out. The MP was preirradiated with 41.5 kGy to stabilize response of the epitaxial diodes. Medium term reproducibility of the preirradiated MP was 2.1%. The MP showed decreased dose per pulse response at higher dose rates (dose per pulse > 1 × 10⁻¹³ Gy/pulse) due to Auger recombination effects. At lower dose rates typical for IMRT deliveries the MP appears to be dose rate independent. MP depth dose measurements agree with ion chamber depth dose measurements within 0.7%. The water equivalent depth of the 50 μm epitaxial silicon diode was found to be 1.9 mm at 1 mm depth in solid water phantom. Dose linearity of the MP was excellent for the measured dose range of 0.05–10 Gy. MP epitaxial diode showed an enhanced response up to a factor of 7.5 at low photon energies that were consistent with other Si devices. MP showed angular response dependency due to the anisotropy of the silicon diode with the maximum angular response of 10.8% at gantry angle 180°. Angular dependence was within 3.5% for the gantry angles ±75°. The field size dependence of the MP at isocenter agrees with the ion chamber measurement to within 1.1%. Field size dependence is higher at the shorter SDD of 58 cm. This may be due to the extra contribution of contaminant electrons by the linac jaws at shorter SDD and geometric “view” of the head components. In the beam perturbation study, the surface dose increased by 7.3% for a 20 × 20 cm² field size and 12.1% for a
30 × 30 cm² field size, at the SDD of 80 cm. For the 20 × 20 cm² field size, the increase of surface dose due to the presence of the MP was 10.9% lower than the COMPASS system. The transmission for the MP was 99% which was 2.3% and 6% higher than other transmission detectors compared in this work. The low beam perturbation and high transmission of the MP shows the suitability of this device to be used for real time fluence map monitoring for IMRT or VMAT treatment.

The magic plate was shown to be a good transmission radiation detector and its dosimetric properties were successfully characterized. Magic plate shows promise as a transmission detector due to the low perturbation of radiation fields arising from the unique new diode packaging in a thin Kapton substrate.

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