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The feasibility study and characterziation of a two-dimensional diode array in “magic phantom” for high dose rate brachytherapy quality assurance

A Espinoza

University of Wollongong, aae718@uowmail.edu.au

B Beeksma

University of Wollongong, bb401@uowmail.edu.au

M Petasecca

University of Wollongong, marcop@uow.edu.au

I Fuduli

University of Wollongong, if473@uowmail.edu.au

C Porumb

University of Wollongong, csp528@uowmail.edu.au

See next page for additional authors

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Abstract

High dose rate (HDR) brachytherapy is a radiation treatment technique capable of delivering large dose rates to the tumor. Radiation is delivered using remote afterloaders to drive highly active sources (commonly ^{192}Ir with an air KERMA strength range between 20 000 and 40 000 U, where 1 U = 1 $\mu\text{Gy m}^2/\text{h}$ in air) through applicators directly into the patient's prescribed region of treatment. Due to the obvious ramifications of incorrect treatment while using such an active source, it is essential that there are methods for quality assurance (QA) that can directly and accurately verify the treatment plan and the functionality of the remote afterloader. This paper describes the feasibility study of a QA system for HDR brachytherapy using a phantom based two-dimensional 11 x 11 epitaxial diode array, named "magic phantom."

Keywords

array, magic, phantom, high, dose, rate, brachytherapy, quality, study, characterisation, two, dimensional, feasibility, diode, assurance

Disciplines

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Authors

A Espinoza, B Beeksma, M Petasecca, I Fuduli, C Porumb, D L. Cutajar, S Corde, M A. Jackson, M LF Lerch, and Anatoly B. Rosenfeld

The feasibility study and characterisation of a two dimensional diode array in “magic phantom” for high dose rate brachytherapy quality assurance

A. Espinoza

Centre for Medical Radiation Physics, University of Wollongong, New South Wales 2522, Australia

5 B. Beeksmā, M. Petasecca, I. Fuduli, C. Porumb and D. Cutajar,

Centre for Medical Radiation Physics, University of Wollongong, New South Wales 2522, Australia

S. Corde and M. Jackson,

Department of Radiation Oncology, Prince of Wales Hospital, New South Wales 2031, Australia

M.L.F. Lerch and A.B. Rosenfeld

10 *Centre for Medical Radiation Physics, University of Wollongong, New South Wales 2522, Australia*

Purpose: High Dose Rate (HDR) Brachytherapy is a radiation treatment technique capable of delivering large dose rates to the tumour. Radiation is delivered using remote afterloaders to drive highly active sources (commonly ^{192}Ir with an air KERMA strength range between 20000 to 40000 U, where $1\text{U} = 1 \mu\text{Gy m}^2/\text{h}$ in air) through applicators directly into the patient’s prescribed region of treatment. Due to the obvious ramifications of incorrect treatment while using such an active source, it is essential that there are methods for quality assurance (QA) that can directly and accurately verify the treatment plan and the functionality of the remote afterloader. This paper describes the feasibility study of a QA system for HDR brachytherapy using a phantom based two-dimensional 11x11 epitaxial diode array, named “Magic Phantom”.

Methods: The HDR brachytherapy treatment plan is translated to the phantom with two rows of 10 (20 in total) HDR source flexible catheters, arranged above and below the diode array “Magic Plate” (MP). Four dimensional source tracking in each catheter is based upon a developed fast iterative algorithm, utilising the response of the diodes in close proximity to the ^{192}Ir source, sampled at 100 ms intervals by a fast data acquisition (DAQ) system. Using a ^{192}Ir source in a solid water phantom, the angular response of the developed epitaxial diodes utilised in the MP and also the variation of the MP response as a function of the source-to-detector distance (SDD) were investigated. These response data are then used by an iterative algorithm for source dwelling position determination. A measurement of the average transit speed between dwell positions was performed using the diodes and a fast DAQ.

Results: The angular response of the epitaxial diode showed a variation of 15% within 360° , with two flat regions above and below the detector face with less than 5% variation. For SDD distances of between 5 and 30 mm the relative response of the epitaxial diodes used in the MP is in good agreement (within 8%) with radial dose function measurements found within the TG-43 protocol, with SDD of up to 70 mm showing a 40% over response. A method for four dimensional localisation

of the HDR source was developed, allowing the source dwell position to be derived within 0.50 mm of the expected position. An estimation of the average transit speed for varying step sizes was determined and was found to increase from (12.8 ± 0.3) cm/s up to (38.6 ± 0.4) cm/s for a step size of 2.5 mm and 50 mm, respectively.

- 40 **Conclusions:** Our characterisation of the designed QA “Magic Phantom” with MP in realistic HDR photon fields demonstrates the promising performance for real-time source position tracking in four dimensions and measurements of transit times. Further development of this system will allow a full suite for QA in HDR brachytherapy and analysis, and for future *in-vivo* tracking.

45 **I. Introduction**

High Dose Rate (HDR) brachytherapy has shown to be beneficial due to its ability to provide a highly localised dose in short treatment times. The advantages and disadvantages of this technique have been well documented within TG-59 report¹. As brachytherapy involves high dose rates (air KERMA strength range between 20000 to 40000 U), minimal fractionations and short treatment times, it is essential that treatments are delivered with high accuracy, as errors will result in severe complications^{2,3}.

While contemporary radiotherapy is very accurate, errors have happened. More than 500 HDR accidents (including one death²) have been reported along the entire chain of procedures from source packing to delivery of dose³. Although most errors in brachytherapy have been the result of human errors, rather than failure of the treatment delivery system and treatment planning, the fast verification of the remote afterloading units and generated treatment plans are important in a “good” quality assurance program. The American Association of Physicists in Medicine ratified that to guarantee a good quality assurance program, the medical physicist should be able to accurately measure the source position, timing, transit velocity and, ideally, absorbed dose distribution⁴.

60 An ideal system for treatment verification should be able to provide the 3D identification of the dwell positions, measure the dwell and transit times, and compare in real-time these parameters with the treatment planning. In the literature there are several studies that investigate different HDR brachytherapy Quality Assurance (QA) approaches. The main goal of the majority of these studies, described below, was to develop a method of localising the source dwell positions and measure timing patterns.

A pinhole imaging system, coupled with an x-ray fluoroscope and radiographic screen-film was developed by Duan *et al*⁵. The system has two pinholes placed away from the patient and is used to reconstruct the source dwell positions in three dimensions using a triangulation reconstruction algorithm. The source was localised within 1 mm and allowed for recording of a HDR treatment by the fluoroscopy machine on to a standard VCR video-tape for post-treatment verification. This method, while able to reconstruct the source position in post-processing of film, relied on the user to notice any significant deviations in position via the fluoroscope screen during actual treatment.

75 Nakano *et al*⁶ proposed the use of diamond detectors for HDR source localisation. The detectors were characterised for their use in HDR, and, using a suggested triangulation method with a minimum of 3 detectors, the source position could be calculated in three dimensions. The intended method would be able to localise the source with an accuracy of 2 mm, according to the authors, within a source to detector distance of 1 to 12 cm, based on the dose to SDD fit used.

80 A method for using a mobile C-arm fluoroscopy machine to image the check cable at dwell positions prior to treatment was investigated by Liu *et al*⁷. Verifying by visual inspection, the authors remarked that the method is useful for observing gross errors, larger than 1 cm, in dwell positions.

The use of a single flat panel detector for tracking the brachytherapy source has been described by Song *et al*⁸. The flat panel is coupled with ball bearings (BBs) in a set pattern on a tray, and uses the shadows of the BBs to determine the coordinates of the source. The source was stepped within a steel applicator, taped to the surface of a plastic slab. The device was capable of determining the position of the source with a mean difference of 0.7 mm with a standard deviation of 1.5 mm. While it was noted that the flat panel used was slow, and required 5 seconds per image acquisition, there are flat panels capable of 45 frames per second and with development into the automation of the BB shadow recognition there is the possibility of fast source tracking.

90 Batic *et al*⁹ expanded the idea of pinhole imaging, using silicon pixelated detectors instead of film. Two catheters were fixed within a support structure with the detection system at a fixed distance away in air. It offers the ability to reconstruct the source position from a source to pinhole distance in air of 40 cm with a precision of about 5 mm. However, this is comparable to the distance between two dwelling positions of the HDR source during treatment and is based on the accuracy of the alignment of the detectors with the pinholes. Although the readout of these detectors allowed for reconstruction within 1 second, giving the possibility of tracking, there was no indication if the detection system was capable of measuring the timing pattern of the source.

100 Rickey *et al*¹⁰ developed a quality check tool for HDR brachytherapy combining the use of radiochromic film and photodiodes for determination of the dwell positions and temporal patterns of the source within the device's fixed treatment tube. Readout of the temporal response of the four photodiodes gave a resolution of 1 ms for measurements of the dwell times. Dwell position are reconstructed with an accuracy of 0.5 mm by manual optical inspection and 0.2 mm by a digital post-processing of the film. The authors state that this method allowed for the verification of the position and dwell times of a single catheter in about 6 minutes. Assuming up to 18 catheters for a treatment plan, the use of this device would not be recommended for QA within clinics.

105 The use of commercial ion chamber arrays, designed for external beam radiotherapy, was investigated by Manikandan *et al*¹¹ for HDR source position verification by placing the catheters directly on top of the device. The I'matriXX (IBA Dosimetry GmbH, Germany) verified treatment dwell positions with a maximum error of 1.8 mm for step sizes larger than 2 cm. It found that due to the large width of each ion chamber (approximately 5.8 mm), the device displayed a volume averaging effect and step sizes below 1 cm could not be resolved.

110

The transit dose of a HDR unit was quantified by Fonseca *et al*¹², based upon calculations of the instantaneous source speed and a Monte Carlo (MCNP5) simulated transit dose profile. The instantaneous source speed, measured using a high resolution optical fiber based detector system¹³ was used to give weighting to the simulated transit dose and allow for an accurate calculation of transit dose values. The results were verified experimentally using radiochromic film and showed that the transit component can be significant, reaching up to a few hundred cGy per application.

Two commercial devices have been developed by Mick Radio-Nuclear Instruments for HDR QA. The PermaDoc GC Phantom¹⁴ allows for the insertion of a single HDR catheter into the device and the source exposes its position as well as centimetre markers onto gafchromic film. This solution verifies one catheter at a time and would require digital post processing to determine an accurate position.

The second device, the MultiDoc Phantom¹⁵, allows for up to 21 catheters inserted into a phantom, where the source can be seen through the catheter, above the phantom. Next to each catheter insert are positional markings in 1 mm increments, and the source dwelling and timing is captured by a high resolution video camera above the phantom. While capable of source position and timing of up to 21 catheters, the system relies heavily on the user to perform verification by watching and analysing the video recording.

These various studies have shown the potential of several systems for HDR QA but may require large and expensive imaging systems and most of them are unable to offer a complete stand-alone real time quality assurance solution, capable of fast and accurate source localisation and analysis combined with high temporal resolution.

This work presents the feasibility study of a new concept of QA for HDR brachytherapy based on “magic plate” (MP) developed at the Centre for Medical Radiation Physics, University of Wollongong, Australia. The MP, a 0.5 mm thick Kapton plate with an embedded two dimensional silicon diode array, was designed for use in a special portable “magic phantom” for fast pre-treatment check of the remote HDR afterloader source dwelling and timing by comparison in real time with the prescribed treatment plan. It could also be used *in-vivo* for intra-operative assessment of the source position, transit time and dwell time within the body of the patient. The MP has a high response to radiation, high radiation hardness, small sensitive volume of each diode for pinpoint measurements, large field of view (10x10 cm²) and readout electronics capable of accurate readout of all diodes within 100 μ s.

This study explores the characterisation of the MP and “magic phantom” for pre-treatment QA system in terms of real time identification of dwell time and source positioning and verification of the transit timing for different distances between dwelling positions. The system is based on a novel adapted for HDR QA 2D diode array, MP, developed at the Centre for Medical Radiation Physics (CMRP),

145 University of Wollongong, which was originally designed as a dosimeter for IMRT/VMAT quality assurance¹⁶.

II. Materials

Several studies have been carried out examining the use of silicon diodes for dosimetry in conventional and intensity-modulated external beam radiotherapy^{17,18,19,20,21}. All these studies agree
150 on the selection of p-type silicon detectors as the best candidate for dosimetry of high energy photons for their high radiation hardness and reproducibility of the response. The MP detector was originally designed for IMRT radiation therapy quality assurance. It comprises of an 11x11 array of epitaxial silicon p-type diodes embedded into a 0.5 mm thick Kapton carrier (Figure 1). Its specialised packaging, using “Drop-in” technology, has been designed to minimise perturbation of the radiation
155 field, avoid dose enhancement due to high Z materials and to maximise the flexibility of the detector to be adaptable to curved shape surfaces such as the human body. MP is 99.8% transparent in a 6 MV photon field when used in fluence mode¹⁶.

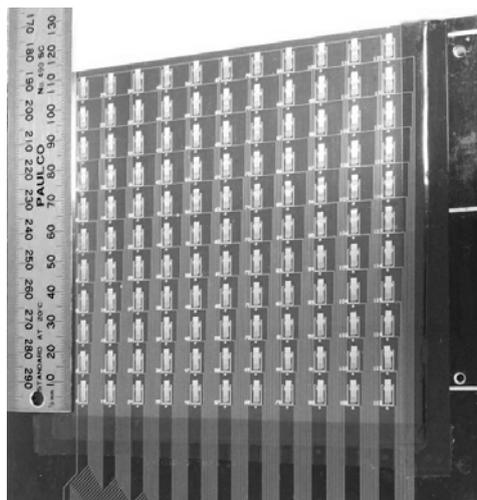


Figure 1 - The magic plate detector array.

160 Each diode is a 50 μm thick p-type (100 $\Omega\text{-cm}$) Si epitaxial layer grown on a 375 μm thick p⁺ (0.001 $\Omega\text{-cm}$) substrate. Dimensions of each detector are 1.5x1.5x0.375 mm³ with 0.6x0.6x0.05 mm³ sensitive volume and 10 mm detector pitch. The MP has been pre-irradiated with ⁶⁰Co gamma photons to stabilise diode sensitivity¹⁶. MP is readout by a multichannel charge to frequency converter preamplifier named TERA06, designed by the Istituto Nazionale di Fisica Nucleare (INFN) – Turin
165 Division and University of Turin microelectronics group for readout of pixelated ionisation or strip chambers for Hadron Therapy²².

The TERA06 is used in conjunction with a Field Programmable Gate Array (FPGA) module. The digital readout and back-end system was designed and developed by the CMRP, and is equipped with

170 a custom designed C++ multi-platform user interface, so named “Rad-X DoseView”. It is capable of reading out each detector in the MP array with zero dead time over a variable acquisition time (between 100 μ s - 100 ms) with real time visualisation and data analysis.

For pre-treatment verification, the MP based phantom system will be able to assess a complete HDR treatment, by verifying the dwell positions and timings, the transit between dwell positions, and the integral dose delivered to the MP diodes for the verified treatment. It allows for comparison with a plan imported from the treatment planning system that is translated to the “magic phantom” and would be capable of real time verification providing identification of error on a particular step of treatment.

180 The MP is sandwiched in a solid water phantom, “magic phantom”, with two insertable rows of 10 catheter holes, up to 20 in total, to suit the needs of the HDR afterloader and prescribed treatment plan as seen in Figure 2. The MP is positioned between the rows of holes and is used to reconstruct in real time the position of the source within the phantom, moving according to the prescribed treatment plan.

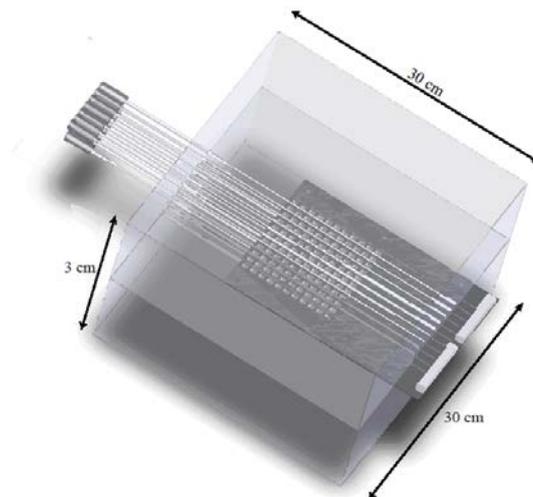


Figure 2 - Illustration of the HDR Pre-treatment QA magic phantom with magic plate inserted.

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Initially, the individual diodes were characterised in terms of their angular response by a ^{192}Ir source in a water equivalent plastic phantom. For these tests, the single epitaxial diodes were mounted onto a 10 mm wide, 25 cm long Kapton carrier with the same packaging process used for MP (“Drop-in” technology). The sensor was readout by a single channel fast electrometer named X-Tream. The X-Tream unit is primarily designed by CMRP as a dosimetry system for Microbeam Radiation Therapy quality assurance²³, and characterised mainly by its high dynamic range ($\sim 10^5$) and high sampling rate (1 MHz).

190

All measurements were taken using the MicroSelectron HDR afterloader (Nucletron, The Netherlands) available at the Prince of Wales Hospital, Randwick (Sydney). The ^{192}Ir source activity varied during the experimental sessions from 40000 U down to 20000 U and all the measurements and planning have been normalised the air-kerma strength of the source. In all the experiments, the HDR source was driven through the transfer tubes, into a flexible plastic catheter.

III. Methods

III.A. Angular Response Measurements

The intrinsic asymmetric structure of the epitaxial diode of MP and the energy range of the gamma photons emitted by the ^{192}Ir may result in directional dependence for each detector (Figure 3). To predict the angular response of the epitaxial diodes in the MP for any position of the source, it is necessary to characterise the angular response of the single diode of the array along both the azimuth and polar angles.

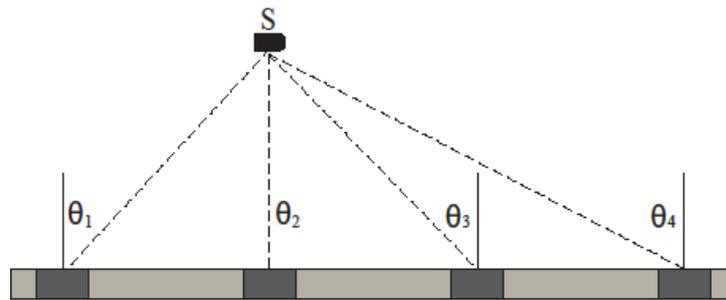


Figure 3 - Variation of angles of MP sensors (dark grey) with respect to HDR Source (black).

To measure the azimuth angular response the diode was placed within a rotatable cylindrical phantom that was encased in a cubic phantom of $30 \times 30 \times 30 \text{ cm}^3$ solid water, to ensure full body scattering conditions. A HDR prostate brachytherapy catheter was placed at 20 mm SDD, as seen in Figure 4. Initial position ($\theta = 0^\circ$) was taken as when the orientation of the detector's sensitive volume was 'face-up' relative to the source.

Using a 28000 U ^{192}Ir source, the MicroSelectron HDR brachytherapy afterloader was programmed to position the source to a distance of 1240 mm (final dwell position at the end of the catheter) for 45 seconds at the dwell position and then retract. The system acquired data at 1 MHz sampling frequency and averaged it over 10 samples to minimise variance due to electrical and thermal noise. The cylindrical insert was rotated in a clockwise direction for a total of 180° rotation with measurements taken at every 30° increment. There was no need to measure along a full rotation due to the rectangular diode design.

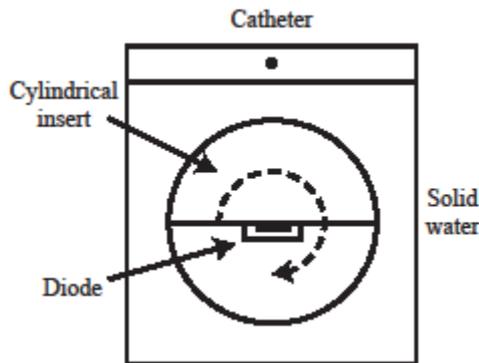


Figure 4 - Cross-section view of the set-up for the measurement of the azimuth detector response. Sensitive volume of the detector in grey.

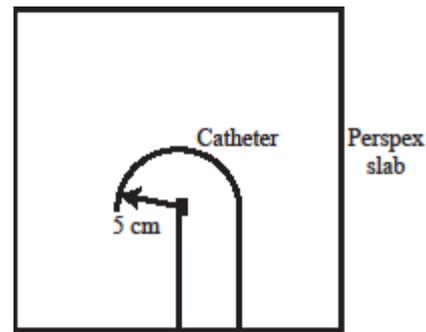


Figure 5 - Cross-section view of the set-up for the measurement of the polar detector response.

Polar angular response characterisation required the design of a specific phantom where the HDR
 220 source would move on an arc of radius 5 cm, centred at the detector to minimise the effect of any
 variation in the SDD and any anisotropy effects of the source. A schematic diagram of the phantom
 is shown in Figure 5. The detector is placed side on in a groove, machined into the phantom. In this
 orientation, when the source is at the final dwell position of the catheter, the detector sensitive volume
 faces the perpendicular axis of the source at 0°. Stepping the source in 13 mm increments around the 5
 225 cm arc radius obtains about 15° angle step. The source was driven around an arc of 180°.

III.B. Source-to-Detector Distance Dose Measurements

Accurate dose verification at regions in the immediate proximity to the source is imperative in HDR
 brachytherapy. Due to the large dose gradients associated with ¹⁹²Ir source, a slight variation in
 position leads to a large variation in dose. Determination of the position of the source using a
 230 triangulation algorithm requires knowledge of accurate relative SDD dose response of the diode
 measured in a water equivalent phantom, with a diode placed face on relative to the HDR source.

The MP was used to measure the response of the diode in a solid water phantom with varying SDDs
 from 6 mm to 71 mm, relative to a stationary HDR source along the central axis of the diode,
 perpendicular to the face of the sensitive volume. The catheter was placed at the centre of a 30x30x30
 235 cm³ solid water phantom and the 28000 U HDR ¹⁹²Ir source was programmed to dwell directly over
 the middle detector of the MP, with the MP placed at different depths in the phantom.

The diode response was normalised to 100% at 15 mm SDD. This SDD was chosen to minimise the
 effect of uncertainty in experimental setup and energy response of the diode associated with the
 softening of the ¹⁹²Ir spectra with SDD²⁴. These results used to calculate the radial dose function and
 240 are compared to the values found within the AAPM TG-43 protocol²⁵. This calculation was
 performed to take into account the 1/r² dependence of the source.

III.C. HDR Source Tracking using magic plate

MP is a 2D array of diodes where each detector element is measuring energy deposited by the gamma photons or secondary electrons emitted by a source with no directional information. However, the radial SDD r can be estimated taking into account the diode response as a function of distance and angle normalised to the activity of the source. Correlation of the measurements taken by three or more detectors can then identify the position of the source in three dimensions using a triangulation method based on estimation and iteration²⁶ (Figure 6). This technique requires the accurate characterisation of the detector response to ¹⁹²Ir gamma spectrum in terms of relative SDD dose and angular responses.

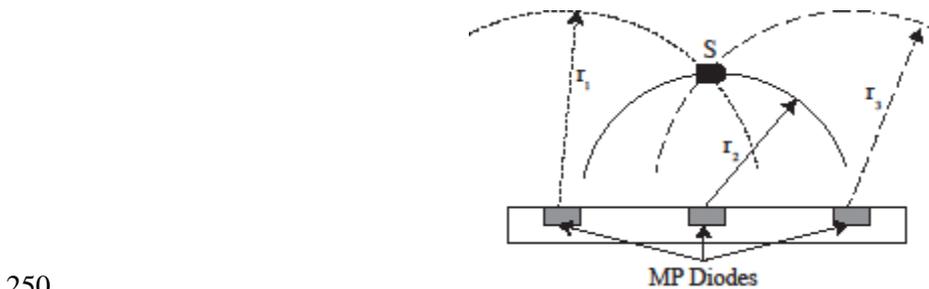


Figure 6 - Triangulation of the HDR source based on the response of three non collinear detectors.

A software tool was developed in C++ to calculate the position of the HDR source in 3D using the real time response of each detector in MP acquired by the TERA system. As three non-collinear detectors are the minimum for localisation of the source, selection of more than three allows for redundancy in estimation of the x, y and z coordinates of the source, and inaccuracy reduction. The MP is an 11x11 array, and every detector can be used for estimating the source position, however, only the detector with the highest response and its immediate neighbours (2x2 or 3x3 subset) were considered to minimise uncertainty due to lower signal from the detectors at larger distances from the source as well as minimise uncertainty with non-isotropic response of the detector and emission of the source. The origin of the detector coordinate system is set at the sensor on the bottom left of MP (Figure 7, point O) that provide coordinate of each detector d_i on MP.

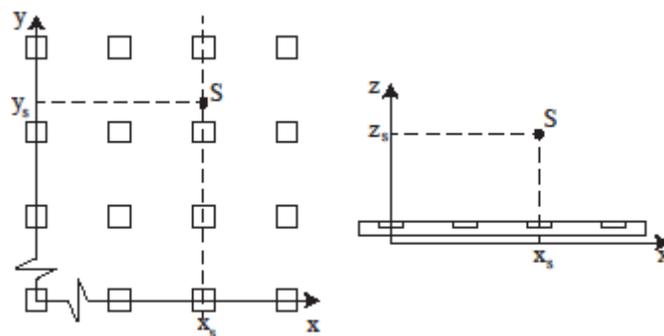


Figure 7 - HDR Source Triangulation co-ordinate system, z axis is out of page. The x_s and z_s are constant with the source travelling only in the y direction.

265 The response of the central diode in the MP per unit source air-kerma strength was measured for
 known activity ^{192}Ir source in water with centre of HDR source located 10 mm above the centre of the
 diode under full backscattering conditions. The MP response was normalised prior to this in a 20×20
 cm^2 photon field from a 6MV LINAC at a source-to-surface distance of 100 cm in a $30 \times 30 \times 30 \text{ cm}^3$
 270 solid water phantom. The MP was placed at a depth of 10 cm, where the photon field has a flat dose
 profile, verified regularly using an ion chambers as part of the LINAC's quality checks.
 Reproducibility of all diodes before normalization was better than 1%. It is assumed that the response
 of the detector varies linearly with the HDR source strength.

The HDR ^{192}Ir source is located at position $S(x_s, y_s, z_s)$. The data acquisition system acquires a frame
 of the whole MP array. The software tool searches for the highest response within the data set. The
 275 detector with the highest response is selected to be D_I and the 8 surrounding detectors not all collinear
 are also selected, when not considering D_I to be on the edges of MP. Using the response of each
 detector, the detectors distance from the centre of the source can be calculated.

The response of the i -th detector, R_i , is a function of its distance r_i from the HDR source, based upon
 the SDD dose response measured, $R_i = f(r_i)$. The distance of the source to the detector i , can be then
 280 calculated by the inverse function $r_i = f^{-1}(R_i)$, derived from the calculation of the radial dose
 function, which was measured for angle $\theta = 0^\circ$. This estimation of distance of the source to the
 detector assumes no angular dependence and is corrected for on further iterations. Based on several
 (at least from three detectors) derived distances r_i the source initial position $S'(x'_s, y'_s, z'_s)$, (Figure 7) in
 the MP frame is estimated.

285 Using the estimated source position, $S'(x'_s, y'_s, z'_s)$, and the coordinate of the i -th detector $D_i(x_i, y_i, z_i)$,
 the geometrical distance d_i is calculated. An iterative technique has been devised to find the source
 position by calculating the minimization of the square of the sums of the percentage difference of the
 values of d_i and r_i with varying position estimations and taking into account angular response of the
 diodes for each consecutive position.

$$\min \sum_{i=1}^n \left(\frac{d_i - r_i}{r_i} \right)^2$$

290 As the source is assumed to be closest to the detector with the highest response, $D_1(x_1, y_1, z_1)$, the sum
 of the squares is calculated using the first estimation of the source position, $S'(x'_s, y'_s, z'_s) =$
 $S'(x_1, y_1, z'_s)$, with z'_s derived from the equivalence of r_1 and the distance between D_1 and the source
 coordinate estimation. The second and third iterations are performed by changing x'_s by ± 5 mm, and
 then determining which sum of the squares is lower. The method is then repeated with a change in the
 295 estimation that is half of the previous, until the change is lower than 0.04 mm. The method is then
 repeated with changes in the estimation of y'_s .

This robust method essentially fits the best estimation of the source position to the detectors response. The relative angle of each diode to the source is then calculated and corrections are then applied to the MP response based upon the angular dependence of each detector. The source position is then re-estimated by the above method.

Validation of the source tracking method was performed using the afterloader programmed to drive the HDR source at 2.5 mm step sizes and with a dwell time of 2 seconds towards the end of the catheter. The catheter was placed within a water equivalent plastic phantom and positioned directly above the middle column of the MP detectors ($x = 50$ mm) with the source stepping across the y direction. The method was evaluated with MP at a SDD of 10 mm below the catheter's plane. The source positions are reconstructed for each acquisition frame of the whole MP using developed routines and compared to the input of the afterloader software.

III.D. HDR source speed measurement using X-Tream System

Due to the fast acquisition sampling rate of the X-Tream system, up to 1 MHz, it is possible to measure the response of a single epitaxial diode (instantaneous dose rate) while the source is in transit. The diode mounted in a kapton tail as mentioned earlier is placed at the centre of a $30 \times 30 \times 30$ cm³ solid water phantom, with the HDR catheter placed at 5 mm distance from the detector.

The HDR source is stepped to the first dwell position within the catheter, and is stationary for 5 seconds. It then moves to the next dwell position, with a measureable time in transit. The source remains at the second dwell position for an additional 5 seconds followed by returning to the afterloader. The step size was varied from 2.5 mm to 50 mm, and the transit time and average speed were recorded. Previous evaluations of the transit velocity using direct measurements have shown a range been 5.4 to 33.1 cm/s^{27,28,29} for the same range of step sizes.

IV. Results

IV.A. Angular Dependence

The directional dependence of the epitaxial diode response is shown in Figure 8. Both data sets are normalised to 100% at 0°. The azimuth data were taken using the cylindrical insert as shown in Figure 4. Determination of the angle for this phantom has a Type A uncertainty of 2.5°. The polar data were taken using the phantom shown in Figure 5. As the source was driven by the HDR loader to set points with 1 mm accuracy, the estimated uncertainty is calculated to be 0.6°. Type A uncertainty in response in both sets is $\pm 0.8\%$ (1 standard deviation), mainly due to electronic noise associated with using a single detector on a kapton tail and is obtained by repeated measurements.

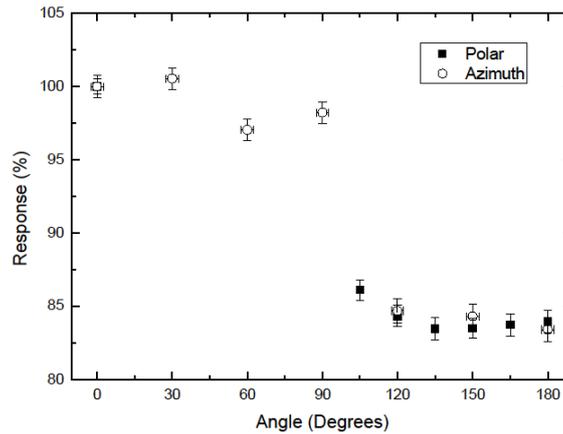


Figure 8 - The angular response of the epitaxial diode (Polar and Azimuth).

330 Both data sets show a variation in response of the diode of 15% between the top face (0°) and bottom
 face (180°) of the epitaxial detector but it is uniform within $\pm 5\%$ of variation for angles range of
 0 $\pm 60^\circ$ and 180 $\pm 60^\circ$. Data measured in the polar direction between 0-110° were excluded due to issues
 in alignment resulting in aberrant values. The obtained angular response in Figure 8 is used in
 modifying the dose response of each detector used within the iterative algorithm for seed position
 335 identification. The regions of flat angular response can be used for better accuracy of seed
 reconstruction algorithm and is corresponding to situation when closest diodes neighbourhood is used
 as in a presented algorithm.

There is an asymmetrical boundary region surrounding the sensitive volume, made up of silicon
 where charge collection is not ideal, leading to a lower amount of charge collected per detected event
 340 in this part of the sensitive volume. The structure of the detector, in conjunction with the energy
 dependence of the response of silicon for the ^{192}Ir source and attenuation of secondary particles
 through the passive silicon, is the cause of the observed variation of the response³⁰.

IV.B. Source-to-Detector Distance Dose Measurements

The SDD dose response of the MP to a ^{192}Ir source in a water equivalent plastic phantom was
 345 measured using the central element of the detector array. The source was positioned directly over the
 sensor at the initial SDD of 6 mm from the central detector and then varied to 71 mm. The type B
 uncertainty associated with the alignment in the z plane was estimated as large as ± 1 mm.
 Measurements over 71 mm were not obtainable by the current readout system due to signal being
 below the measurable threshold.

350 Based on the TG-43 Protocol, a calculation of the radial dose function was performed using the MP
 SDD dose response, taking into the account the estimated radial distance and the ^{192}Ir source physical
 dimensions, shown in Figure 9. This calculation takes into account the $1/r^2$ dependence and allows for
 deviations from the theoretical to be clearly seen. The large uncertainties shown within the figure at

low SDDs are due to the combination of the uncertainty of the repeated measurements and the large percentage error from using a 1 mm positioning uncertainty.

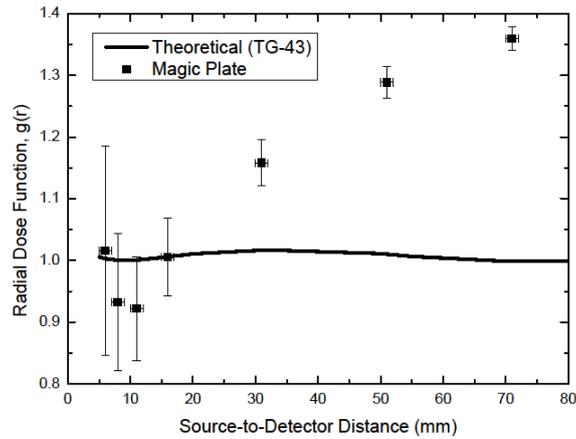


Figure 9 - Comparison between the radial dose function, $g(r)$ for ^{192}Ir from the TG-43 protocol and that based upon the MP response.

Deviations from expected response are seen at source-to-detector distance over 15 mm. The measured followed the TG-43 calculated radial dose function up to 30 mm within 8% and for 70 mm within 40%, respectively. This is due to a possible over response of the silicon detector at these SDDs due to a larger lower energy scatter contribution³¹. Although there is a differing response of MP compared to the TG-43 protocol, it does not interfere with the source position determination accuracy proposed in this work.

IV.C. Evaluation of HDR Source Tracking

Comparison of the positions and dwell times determined by the HDR source tracking algorithm and those inputted into the Nucletron microSelectron HDR afterloader computer show good agreement for the MP SDD of 10 mm as seen in Figure 10. Dwell times are all measured to be 2 s, with the catheter position measured to be on average $x_s = 50.3 \pm 0.1$ mm and $z_s = 10.5 \pm 0.4$ mm with type A uncertainties calculated by 1 standard deviation.

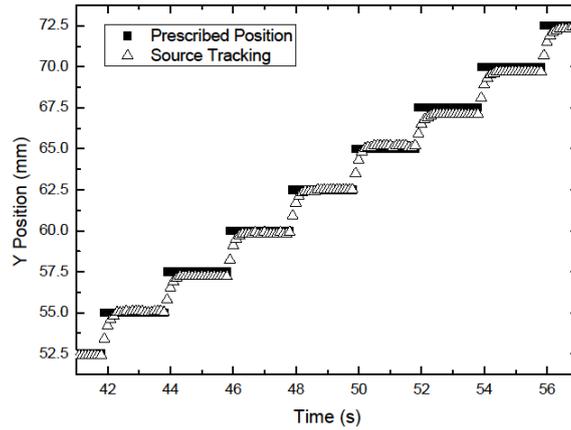
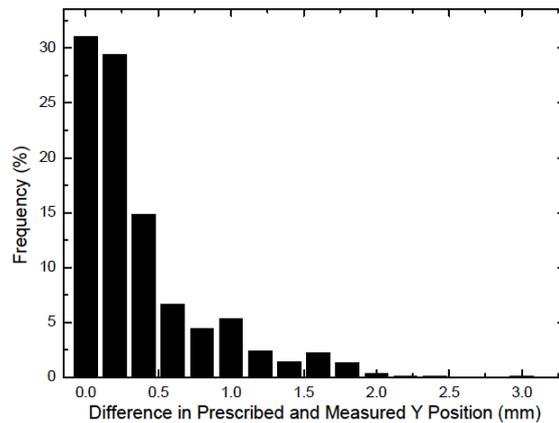


Figure 10 - Comparison of source tracking calculation of a HDR source of step size 2.5 mm and dwell time of 2 seconds against expected prescribed position ($z_s = 10$ mm)

375 It is also possible to see a slight drop in expected position at the start of the dwell times. This is due to the source being measured in transit between dwell positions with an acquisition speed of 100 ms. Using a faster acquisition frequency, it is possible to accurately measure the source in transit, and determine the source transit time.



380 Figure 11 - Frequency histogram showing the difference in the prescribed position and that measured using the MP source tracking algorithm for the described treatment plan.

Figure 11 shows the frequency distribution of the difference between the prescribed position and the measured position of the HDR source with approximately 75% of measurements falling within 0.5 mm. The occurrences of the larger differences are due to the source in transit. This result demonstrates the feasibility to track the HDR source in real time, accurately, over a wide field of view of MP (10x10 cm²).
385

IV.D. High Speed HDR Source Tracking using X-Tream System

Figure 12 shows the temporal response of the single epitaxial diode using the X-Tream acquisition system. The HDR source moves through the catheter to the first dwell position, where it is stationary

for 5 seconds, then moves 15 mm to the final dwell position, and is motionless for an additional 5
390 seconds. The source finally retracts back into the afterloader after this time. An increase in response
can be seen as the second dwell position is closer to the detector. Measurements were taken over 16
seconds, at 1 MHz sampling rate; data has been averaged by a factor of 10 to improve the signal to
noise ratio on the measurements and to make data handling and processing easier. Standard deviations
of the average system response for one measurement in the first and second dwell positions are found
395 to be 3.9% and 1.9%, respectively.

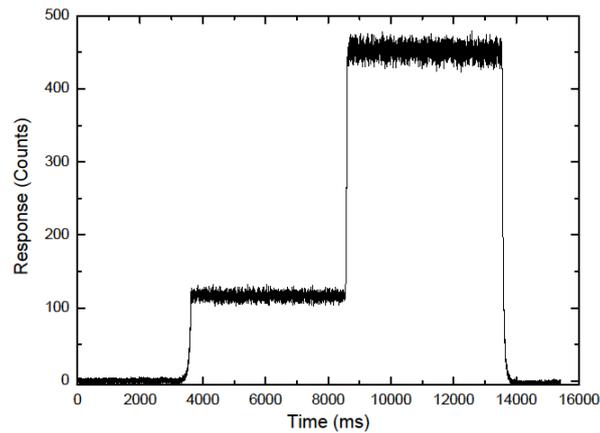


Figure 12 - Temporal response of epitaxial diode due to HDR source being stepped toward diode with 15 mm step size (two dwelling positions).

Due to the high speed of the acquisition system, there are approximately 5000 samples when
400 measuring the HDR source in transit, as seen in Figure 13. The source transit time was measured as
the time interval between the two events when the detector response is changing more than one
standard deviation above the average response at the two consecutive dwell position, to when the
source reaches the average value of response of the second dwell position, thus giving an average
speed for a 15 mm step size. The average transit speed was evaluated as a function of step size in the
405 range between 2.5 and 50 mm for 3 measurements per step size.

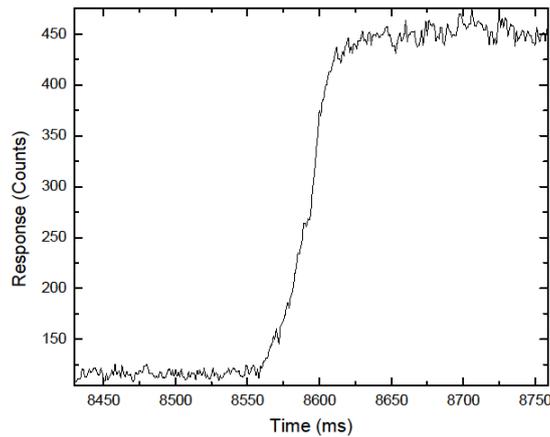
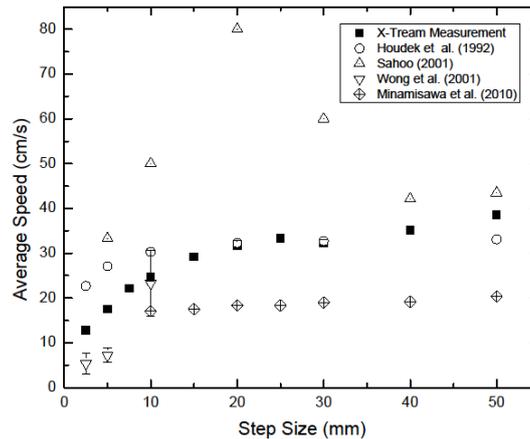


Figure 13 - Magnified view of averaged response due to HDR source movement between dwell positions for a 15 mm step size.

The results are shown in Figure 14, along with values published. The type A uncertainty of each result is less than 5% and is calculated based on the standard deviation of the average speed of three measurements per step size. Values found within the literature have a large variation, showing the challenging nature of this characterisation. Measurements performed by Sahoo²⁷ using an Ion chamber recorded up to 100% uncertainty due to the poor timing resolution of the system. The X-Tream measurement shows good agreement with the measurements performed by a direct estimation of the transit time in Houdek *et al*²⁹, which was performed using an oscilloscope to measure the pulse generator of the stepping motor of the afterloader; also the data from Wong *et al*²⁸ obtained by the use of a high speed camera shows good agreement. Minamisawa *et al*¹³ used an optical fibre system to detect the radiation generated by the moving HDR source to calculate the instantaneous and average velocity.

Although the discrepancy of absolute value is related to the specific afterloader, this direct measurement technique show similar dependence of the source velocity with the step size. The large volume of the detector and indirect estimation of the speed lead to significant errors and uncertainty especially for short dwell distances in this method.



425 Figure 14 - Average transit speed measured using X-Tream System (black). Results extracted from the literature are shown
for comparison.

Although there is good agreement with previous results, the use of a single detector is not recommended for an accurate evaluation of the source speed as only average source speed is measured. Using the MP array, with its constant detector spacing and a sampling rate of 100 μ s,
430 instantaneous speed measurements could be performed for any positions within treatment related volume, similar to the method proposed by Minamisawa *et al*¹³, which uses an optical fibre system and digital oscilloscope with timing resolution of 1 nanosecond.

V. Conclusion

The MP detectors, detector array and readout electronics, designed originally for use in IMRT and
435 VMAT radiotherapy quality assurance, were characterised for their use in fast HDR QA. The developed epitaxial diodes, which are attractive due to their radiation hardness and high stability, were characterised in terms of measured dose and angular responses using the ¹⁹²Ir HDR source. Further development will allow for reliable measurements at larger distances, necessary for future in-vivo source tracking applications and in-vivo dosimetry based on known location of the source,
440 relative to anatomy.

Based on the detailed characterisation of “Magic Phantom” for QA in HDR brachytherapy using ¹⁹²Ir HDR source, the selection of the distance between MP and upper and lower planes of catheters 5-10 mm away ensures the accuracy of the iterative method. The “Magic Phantom” for QA in HDR brachytherapy will verify the correctness of the plan delivery, catheter by catheter in terms of
445 dwelling position and time, and will help to avoid accidents associated with the malfunction of the afterloader. A comparison of the “Magic Phantom” and selected devices from the literature for HDR quality assurance is presented in Table 1.

Table 1 - Comparison of "Magic Phantom" with selected systems found with literature.

	"Magic Phantom"	Radiochromic Film and Photodiodes ¹⁰	Commercial Ion Chamber Array ¹¹	Mick® MultiDoc™ Phantom ¹⁵
Can test full treatment plan?	- Yes, currently up to 20 catheters.	- Yes, but only one catheter per test. Not in real time	- Tested with 3 catheters only	- Yes, up to 21 catheters.
Dwell Position measurement uncertainty/error:	- 75% of source positions were measured within 0.5 mm of expected position and in real time.	- 0.5 mm by manual inspection, 0.2 mm by digital off line processing.	- 1.01 mm standard deviation.	- Positional Markings in 1 mm increments.
Can check dwell times?	- Yes.	- Yes.	- No.	- Post image analysis is required.
Timing Resolution:	- Currently 1 ms.	- 1 ms.	- Unknown	- Timing resolution of video camera.
Advantages:	- Instant Readout of results via USB PC software tool. - Inexpensive device.	- Direct measurements of stationary source locations. - Film which can be kept for records.	- Exists in clinical setting. - Water equivalent detectors.	- High resolution video recorded for records.
Disadvantages:	- Currently not focused on dosimetric information.	- Needs expensive Film for every test. - Time consuming to verify full plan.	- Need for post processing of data. - Limited to source step size >1cm. - Expensive device.	- User needs to watch and analyse video recording.

450 By reconstructing the source position based on directly measuring the radiation source any incorrect estimate of the source position relative to the needle tip in planning will also be immediately identified during QA of the treatment plan. It can be used for full plan QA or for a partial verification for accelerated QA for each patient plan. A developed software interface will allow for the immediate

455 visualisation of results and analysis of acceptance parameters in terms of tolerance of actual dwelling
positions and times, in comparison with the treatment plan.

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460 Address of the corresponding author: anatoly@uow.edu.au

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