NOTE

Evaluation of the Delta^4 phantom for IMRT and VMAT verification.

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Abstract

The Delta^4 diode array phantom (Scandidos, Uppsala, Sweden) was evaluated for verification of segmental intensity-modulated radiation therapy (IMRT) and volumetric modulated arc therapy (VMAT) on an Elekta linear accelerator (Crawley UK). The device was tested for angular sensitivity by irradiating it from 36 different gantry angles, and the responses of the device to various step-and-shoot segment doses and dose rates were evaluated using an ionisation chamber as a comparison. The phantom was then compared against ionisation chamber and film results for two prostate and pelvic nodes IMRT plans, two head and neck IMRT plans and two lung VMAT plans. These plans were calculated using Pinnacle^3 (Philips Radiation Oncology Systems, Madison, WI). The uniformity of angular response was better than 0.5% over the range of gantry angles. The uniformity of response of the Delta^4 to different segment monitor units and dose rates was better than 0.5%. The assessment of the IMRT and VMAT plans showed that the Delta^4 measured a dose within 2.5% of the ionisation chamber, and compared to film recorded a slightly larger region (range –2% to +7%) agreeing with the planned dose to within 3% and 3 mm. The Delta^4 is a complex device and requires careful benchmarking, but following the successful completion of these measurements, Delta^4 has been introduced into clinical use.

1. Introduction

Complex radiotherapy treatment plans such as those involving intensity-modulated radiation therapy (IMRT) and volumetric modulated arc therapy (VMAT) require dosimetric verification before clinical delivery (Ezzell *et al* 2003). The Delta^4 phantom (Scandidos, Uppsala, Sweden) offers a convenient means of achieving this. It consists of 1069 p-type Silicon diodes in a crossed array inside a cylindrical polymethylmethacrylate (PMMA) phantom, and associated computer software allows the user to compare the measured dose distribution for a complete treatment plan with the dose distribution predicted by the treatment planning system. The diodes are cylindrical, have an area of 0.0078 cm^2, and are spaced at 0.5 cm intervals over the central 6 x 6 cm of the planes and at 1 cm intervals over the remainder of the central 20 x 20 cm of the planes. The crossed planes are achieved by means of a main detector board which passes through the entire diameter of the phantom, and two wing detector boards which are separated to allow the main detector board to pass between them. The phantom itself has a diameter of 22 cm and length of 40 cm (Figure 1). The device records measured dose in relation to the individual accelerator pulses by using a trigger signal from the accelerator, thereby facilitating time-dependent four-dimensional applications. Gantry angle is independently sensed by means of an inclinometer attached to the gantry or accelerator head. This allows the device to identify which control point of a dynamic arc delivery is being delivered, so that the measured dose can be associated with this control point, and the appropriate correction for gantry angle applied.

The verification process requires the treatment plan in consideration to be recalculated on a CT scan of the phantom. An artificial dataset consisting of a uniform PMMA-equivalent cylinder is used rather than a real CT scan, so as to avoid errors due to the appearance of the diodes on the CT scan. Calibration of the phantom once a year is recommended. A three-dimensional dose option is available which takes the measured dose in the planes and then rescales the planned dose accordingly to give an indication of the dose that would be measured between the two detector planes. This has not been used in this study as it is intended for interpretation and visualisation of results rather than as a direct measurement.

As this is a new device with new technology, an evaluation and benchmarking process is desirable before clinical use. Furthermore, since the device is not strictly independent of the accelerator, due to the trigger signal obtained from the accelerator, it is important to ensure that the device functions correctly for the accelerator in use. This study therefore evaluates the phantom by means of basic performance tests and

by comparing its results with those produced by the more established methods of ionisation chamber and film (Bedford *et al* 2008b). Segmental (step and shoot) IMRT and VMAT are specifically considered, on Elekta linear accelerators.

2. Methods

2.1. Performance tests

Several basic tests were carried out to examine specific performance characteristics of the Delta^4 phantom. These were not intended to be an exhaustive characterisation of the device, but rather to evaluate those features of the phantom in which greater confidence was required. The phantom was calibrated relatively and absolutely for a 6 MV beam. The November 2008 version of Delta^4 software was used for all tests, and the work was carried out on an Elekta Synergy accelerator with a Beam Modulator head. This accelerator was running RTDesktop v7.01 and was fully commissioned for IMRT and VMAT (Hansen *et al* 1998, Bedford and Warrington 2009). The couch top was a carbon-fibre iBeam EVO type (Medical Intelligence (Elekta), Schwabmünchen, Germany) and the measurements were carried out approximately 0.5 m from the end of the full width section.

2.1.1. Angular sensitivity

The diodes in the phantom have an inherent angular sensitivity to radiation. This is corrected for by the Delta^4 software, which compensates according to the known gantry angle of the beam being delivered. In order to check the accuracy of this correction, 36 10.4 x 9.6 cm beams with 200 MU were delivered at 10° gantry intervals around the phantom. (10.4 x 9.6 cm on a Beam Modulator head is the equivalent of a 10 x 10 cm field on a normal head.) In order to measure the response of the phantom without the attenuation of the couch, the measurements were taken with the phantom in three positions. Firstly, with the phantom upright in its normal position, the responses of the Delta^4 to gantry angles 270° to 90° were measured. The phantom was then turned first onto one side and then onto the other, and gantry angles 270° to 90° used to irradiate from the directions corresponding to 90° to 270° through 180° in normal operation.

The response of the phantom was measured using the daily output correction facility of the Delta^4. The daily output correction measures the overall response $\mathbf{D}_{D4}, \left\lfloor D_{D4}^{1} \ldots D_{D4}^{N} \right\rfloor$ of the N=299 central detectors

in the Delta^4 to a simple field, such as a 10 x 10 cm field, and offers the correction factor needed for agreement with the corresponding planned dose distribution, $\mathbf{D}_{TPS}$, $\left[D_{TPS}^1 ... D_{TPS}^N\right]$:

$$DCF = \sum_{i=1}^{N} D_{TPS}^i \Big/ \sum_{i=1}^{N} D_{D4}^i \,. \qquad (1)$$

This is normally used for setting up the phantom according to the daily output of the accelerator. However, for a fixed planned distribution, $\mathbf{D}_{TPS}$, the relative response, R, of the Delta^4 is given by:

$$R = \frac{1}{DCF}, \qquad (2)$$

which was used in this case for evaluation of the angular behaviour.

Four measurements were taken at each gantry angle to assess the reproducibility and uncertainty. Ideally, the output of the accelerator should have been simultaneously measured with an ionisation chamber to ensure that any variation in Delta^4 output was entirely due to the device itself and not due to accelerator output variation, but due to the variation in gantry position, this was not feasible. However, the response of the phantom to irradiation from the cardinal angles was measured with the phantom both upright and lying down. (For example, the response to irradiation from the top of the phantom was measured with gantry angle 270° and the phantom lying down on its left side, with gantry angle 0° and the phantom standing up, and with gantry angle 90° and the phantom lying down on its right side.) This provided a measure of the uncertainty resulting from the phantom positioning and possible variation of the accelerator with gantry angle.

2.1.2 Linearity of segment dose

An earlier version of Delta^4 software was found to be inaccurate for multiple low-MU IMRT segments. The linearity of the Delta^4 for delivery of segmental beams was therefore checked. A series of 10.4 x 9.6 cm beams were delivered at gantry angle 0°. The beams were for one segment with 80 MU, two segments with 40 MU each, four segments with 20 MU each, eight segments with 10 MU each, 16 segments with 5 MU

each and 32 segments with 2.5 MU each, covering the range of segment sizes encountered in IMRT delivery. The response of the phantom, *R*, was measured using the daily output correction factor as in section 2.1.1 above. A 0.6 cm^3 Farmer ionisation chamber (Saint Gobain Crystals and Detectors, Reading, UK) was positioned under 3 cm of Solid Water (Radiation Measurements, Inc., Middleton, WI) centrally beneath the Delta^4 phantom so as to correct for any variation in accelerator output, which might occur at the small segment doses. For an ionisation chamber charge of *C*, the true response of the Delta^4, R_C, excluding any accelerator variation, was taken as:

$$R_C = \frac{R}{C}. \qquad (3)$$

Four measurements were taken for each beam to assess the uncertainty in the measurement. The value of R_C was calculated for each measurement, and the mean taken as the overall Delta^4 response.

2.1.3. Dose rate dependence

The output of the Delta^4 was examined as a function of dose rate. A series of 10.4 x 9.6 cm beams were delivered at gantry angle 0°. The dose rate of the beam was set to 600 MU/min, 300 MU/min, 150 MU/min, 75 MU/min and 37 MU/min. The output of the Delta^4 was measured using the daily output correction factor. A Farmer chamber was also used to simultaneously correct for the accelerator output as in 2.1.2 above. Ion recombination within the ionisation chamber was measured at 600 MU/min and 37 MU/min and found to be around 0.5% in both cases, which was expected, as the dose per pulse of the Elekta accelerator is constant at all dose rates. Ion recombination was determined to be a constant factor for the dose rate dependence measurements and was subsequently neglected. Four measurements were taken for each beam to assess the uncertainty in the measurements.

2.2. Comparison studies

Having established that the basic performance of the Delta^4 was satisfactory, the device was compared with the more traditional IMRT plan verification methods using ionisation chamber and film in a water-equivalent phantom. Six treatment plans were considered: two prostate and pelvic nodes IMRT plans, two head and

neck IMRT plans and two lung VMAT plans. The prostate and pelvic nodes plans consisted of five segmental (step and shoot) beams, with approximately 10 segments per beam (Adams *et al* 2004), while the head and neck plans consisted of seven beams with approximately 10 segments per beam. The VMAT plans each consisted of a single gantry arc (Bedford *et al* 2008a). The IMRT plans were computed on the Pinnacle^3 treatment planning system (Philips Radiation Oncology Systems, Madison, WI), and the VMAT plans were generated using an in-house program AutoBeam and a final dose calculation was made within Pinnacle^3. The Pinnacle^3 treatment planning system had previously been fully commissioned for IMRT (Bedford *et al* 2003, 2004). As the use of the planning system with the Delta^4 involved calculating dose in PMMA, some additional measurements of dose for various depths and field sizes were made to ensure that Pinnacle^3 was performing this correctly.

The six treatment plans were verified using an ionisation chamber and film, prior to patient treatment. The plans were recalculated on a CT scan of a stack of Solid Water, without changing the monitor units. The expected ionisation chamber dose, $D_{TPS,\,CUBE}$, was calculated by taking the mean dose over a small cylindrical volume representing the ionisation chamber. The plans were then delivered using an Elekta Synergy accelerator with standard multileaf collimator head and RTDesktop v7.01, and daily output of the accelerator was corrected for in all cases. The ionisation chamber measurements were made with a 0.6 cm^3 Farmer chamber in a rectangular stack of Solid Water in a region of homogeneous dose. The ionisation chamber difference, ΔD_{IC}, was then calculated as:

$$\Delta D_{IC} = \frac{D_{IC,CUBE} - D_{TPS,CUBE}}{D_{TPS,CUBE}} \times 100\% , \qquad (4)$$

where $D_{IC,\,CUBE}$ was the dose measured by the ionisation chamber, corrected for the daily output of the accelerator. The value of ΔD_{IC} was required to be such that $-3\% \le \Delta D_{IC} \le +3\%$.

The film measurements were made using whole sheets of 30.5 cm x 25.4 cm EDR2 film (Eastman Kodak, Rochester, NY), with the exception of one of the head and neck cases and one of the lung cases (cases 3 and 6 respectively), where the film measurements were made with whole sheets of 25.4 cm x 20.3 cm Gafchromic EBT film (International Specialty Products, NJ). The film was located coronally in the same Solid Water stack as the ionisation chamber. The EDR2 film was calibrated by irradiating a separate sheet

with the depth dose from a 10 x 10 cm field, and the films were digitised using a Dosimetry Pro Advantage film scanner (Vidar Systems Corporation, Herndon, VA). The EBT film was calibrated using a batch calibration and each film was scanned before and after irradiation using an Expression 1680 Pro flatbed scanner (Seiko Epson Corporation, Nagano, Japan). The films were analysed using OmniPro I'mRT (Wellhöfer-Scanditronix, Schwarzenbruck, Germany) and normalised to a region of homogeneous dose. The gamma index ($\boldsymbol{\Gamma}$) for 3% of the dose to the normalisation point and 3 mm was calculated over a rectangular region of interest positioned around the high-dose region (Low *et al* 1998). This was a two-dimensional gamma calculation, in which the information in the film plane was compared with the information in the corresponding plane of the calculated dose. For each film, 90% of the region of interest was required to have a gamma value of less than unity.

For the Delta^4 evaluation, the plans were recalculated on the artificial CT scan of the Delta^4 without changing the monitor units, giving the dose distribution $\mathbf{D}_{TPS,CYL}, \left[D^1_{TPS,CYL} \ldots D^N_{TPS,CYL}\right]$ in the cylindrical Delta^4 phantom geometry. The output of the accelerator was measured using an ionisation chamber and the Delta^4 daily output correction factor was set to compensate for any deviation from calibration conditions. The plans were then delivered to the Delta^4 phantom and the response compared with the ionisation chamber and film measurements. The absolute dose difference ΔD_{D4} was calculated by making use of the daily output correction factor suggested by the Delta^4 for the delivered plans (see equation (1)):

$$\Delta D_{D4} = \left(\frac{1}{DCF} - 1\right) \times 100\% \quad = \quad \frac{\sum_{i=1}^{N} D^i_{D4,CYL} - \sum_{i=1}^{N} D^i_{TPS,CYL}}{\sum_{i=1}^{N} D^i_{TPS,CYL}} \times 100\% \qquad (5)$$

where $\mathbf{D}_{D4,CYL}, \left[D^1_{D4,CYL} \ldots D^N_{D4,CYL}\right]$ was the dose distribution measured by the Delta^4. The difference ΔD_{D4} therefore reflected all of the central diode measurements around the isocentre.

The overall response was assessed by selecting the diodes measuring greater than 20% of the isocentre dose, and recording the percentage of these diodes which agreed with the plan to within 3% of isocentre dose and 3 mm. This gamma calculation compared the dose at the discrete diode locations with the three-dimensional dose distribution calculated by the treatment planning system.

3. Results

3.1. Performance tests

3.1.1. Angular response

The angular response of the Delta^4 at a range of gantry angles is shown in Figure 2. The response shown is the mean of four measurements at each angle. The variation of the response with gantry angle is less than ±0.5%. The reproducibility of the four measurements at each gantry angle is better than ±0.05% so is not shown in the figure. However, the variation of the response at the cardinal angles with the phantom repositioned and irradiated with a different gantry angle, is ±0.4%, indicating an uncertainty in the measurements of ±0.4%. This is indicated in Figure 2.

3.1.2. Linearity of segment dose

The responses of the ionisation chamber and Delta^4 as a function of IMRT segment dose and number are shown in Figure 3. The response of the ionisation chamber is taken to represent the output of the accelerator. With segments of 5 MU or larger, the variation in both accelerator output and Delta^4 response is better than ±0.5%. With segments of 2.5 MU, the mean accelerator output is within 0.5% of the output at 80 MU, but the output has a range of –1.7% to +1.0%. The corresponding mean Delta^4 response, excluding the accelerator variation, is within 1.0% of the response at 80 MU, with a range in response of –2.2% to +0.5%.

3.1.3. Dose rate dependence

The response of the ionisation chamber and Delta^4 as a function of accelerator dose rate is shown in Figure 4. There is a tendency for the accelerator output to drop and for the Delta^4 to under-respond at lower dose rates, but these are small effects. The accelerator output is within 1.1% of that at 600 MU/min at all of the dose rates studied, and the Delta^4 output, excluding the accelerator variation, is within 0.4% of its response at 600 MU/min.

3.2. Comparison studies

The basic tests of planning system accuracy show that Pinnacle[3] calculates dose in PMMA to within ± 2% for simple square fields.

An example of the Delta[4] output for one of the head and neck cases is shown in Figure 5, and the comparison of Delta[4] with ionisation chamber and film for all of the clinical cases is shown in Table 1. There is no particular trend in the differences between doses recorded by ionisation chamber and Delta[4], but the agreement is to within ±2.5%. On average, the Delta[4] records a higher percentage of points agreeing with the planned dose to within 3% and 3 mm than film.

4. Discussion

The Delta[4] at first appears to be a straightforward device for measuring dose. However, because of the connection to the accelerator trigger output, which means that it is not necessarily completely independent of the accelerator, and because of the various software corrections made to the basic diode response after measurement, the device is complex and careful quality assurance and benchmarking before use is therefore recommended. In particular, the treatment plan to be verified must be recalculated on an artificial CT scan of the Delta[4] phantom, which has a higher density than water. This density is less than optimal for most treatment planning systems. In the case of Pinnacle[3] used in this study, it is known that the dose is subject to some uncertainty at high tissue density (Bedford *et al* 2003). Further measurements carried out specifically in PMMA have shown that the dose calculated by Pinnacle[3] for simple fields on PMMA is accurate to within ±2%.

The performance tests carried out in this study show that the basic Delta[4] behaviour is accurate. The angular response is uniform over the complete range of gantry angles. Couch attenuation is a very much larger factor than the inherent Delta[4] response, varying from 2-5% depending upon the obliquity of the beam in relation to the couch top. For a VMAT plan, where roughly one third of the arc is through the couch, the iBeam EVO couch attenuation causes a drop in overall central dose of around 1%. The effect of the couch should therefore be modelled in the treatment planning system, although this is often not a straightforward process on a routine basis. The response of the Delta[4] for segmental IMRT beams is accurate to within 0.5%, and the device handles the variable dose rate required for VMAT properly.

For the plan comparisons, the Delta[4] absolute response is within about 2.5% of the ionisation chamber measurements. However, as the ionisation chamber measurement is over a much smaller volume

than the Delta^4 measurement, perfect agreement is not expected. Couch attenuation is present on both the ionisation chamber and Delta^4 measurements. The gamma measurements are in reasonable correlation, with the cylindrical Delta^4 showing slightly better agreement between measured and calculated dose than the film in the cuboid phantom. There are several factors which may influence this agreement. In particular, Delta^4 measurements are absolute, whereas film is used in this work as a relative dosimeter. Absolute measurements with film are possible, but as such measurements require a high level of quality assurance in the calibration, processing (in the case of radiographic film) and readout, this is difficult in a routine setting. Consequently, an absolute dose difference manifests itself in the Delta^4 gamma, whereas such a difference is typically removed in the film gamma by normalisation. On the other hand, the Delta^4 is not susceptible to variation in handling like film. Moreover, Delta^4 uses a three-dimensional gamma calculation, which is less challenging than the two-dimensional one used for film. Due to these factors, the results of the Delta^4 should not necessarily be tested against the same criteria as the results of film and ionisation chamber. Based on the results of these tests, it seems reasonable to maintain a tolerance of 90% of the gamma map achieving 3% and 3mm agreement, or perhaps tighten the tolerance slightly to 95% of the gamma map achieving 3% and 3mm.

Following these successful comparative measurements, the Delta^4 has been introduced into clinical use for IMRT and VMAT verification.

Acknowledgments

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Figure captions

Figure 1. The Delta^4 phantom and associated devices in position on the accelerator couch.

Figure 2. Delta^4 response as a function of gantry angle, normalised to the response at gantry angle 0°. The solid line shows the mean of four measurements at each gantry angle, and the broken lines show the estimated uncertainty of the measurement.

Figure 3. Variation of Delta^4 response, as a function of number of segments and segment monitor units. The circles show four ionisation chamber readings, and the broken line gives their mean, representing accelerator output. The squares show four Delta4 measurements, excluding variation in accelerator output, and the solid line gives their mean. Normalised to 1 x 80 MU response.

Figure 4. Variation of Delta^4 response as a function of accelerator dose rate. The circles show four ionisation chamber readings, and the broken line gives their mean, representing accelerator output. The squares show four Delta4 measurements, excluding variation in accelerator output, and the solid line gives their mean. Normalised to 600 MU/min.

Figure 5. Delta^4 output for one of the head and neck cases. Planned dose distribution in greyscale and measured dose in colour over (a) the wing detector boards, and (b) the main detector board. (c) Histogram of 3% and 3 mm gamma values for the measurements in relation to the treatment planning system.

Tables

Table 1. Comparison of ionisation chamber, film and Delta^4. The ionisation chamber difference is the percentage difference of the measured dose in relation to the dose calculated by the treatment planning system. Delta^4 difference is the percentage difference of the measured dose in relation to the dose calculated by the treatment planning system, as indicated by the daily output correction factor. Gamma (Γ) results are the percentage of points passing the gamma criterion of 3%/3mm.

Case	Plan type	Ion chamber diff (%)	Film Γ (%)	Delta^4 diff (%)	Delta^4 Γ (%)
1	**IMRT prostate/pelvic nodes**	-0.4	89.2	-2.3	96.1
2	**IMRT prostate/pelvic nodes**	-1.0	98.2	-1.6	98.7
3	**IMRT head/neck**	-1.0	91.5	1.2	97.7
4	**IMRT head/neck**	-0.6	93.2	0.2	96.5
5	**VMAT lung**	-1.1	97.6	-2.2	95.7
6	**VMAT lung**	2.3	91.5	0.1	98.8

Figures

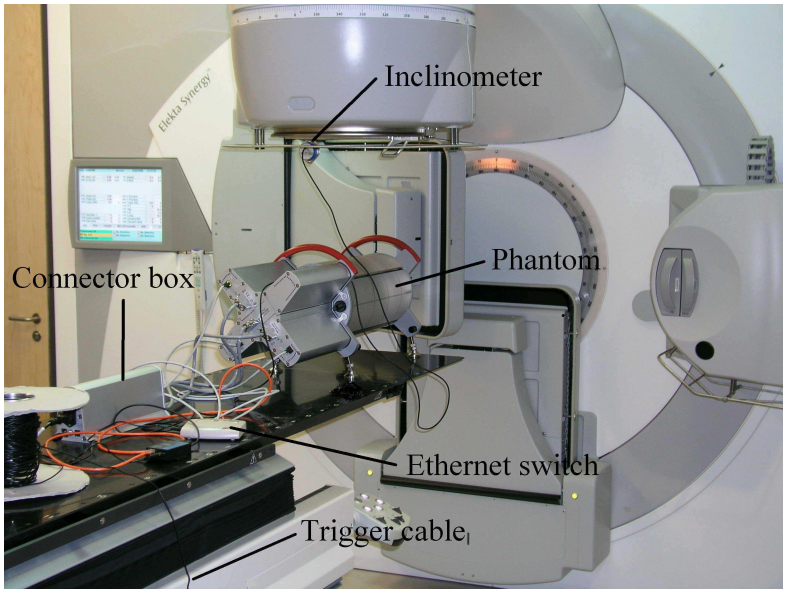

Figure 1.

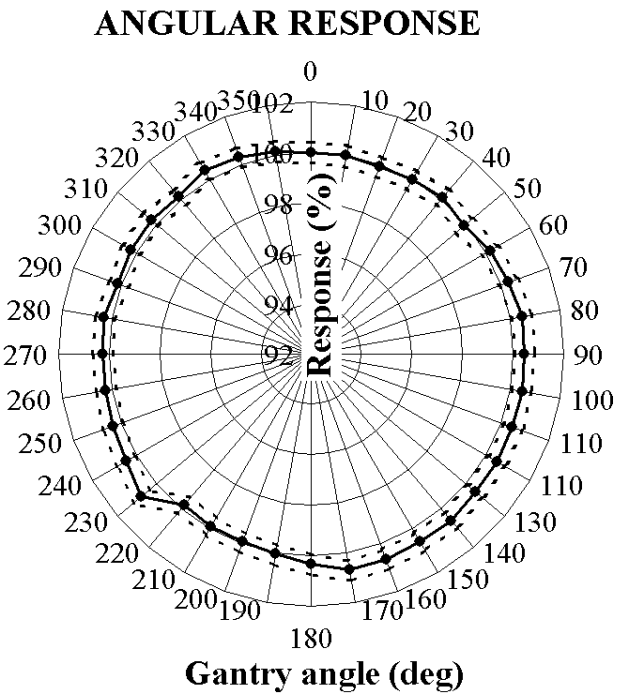

Figure 2.

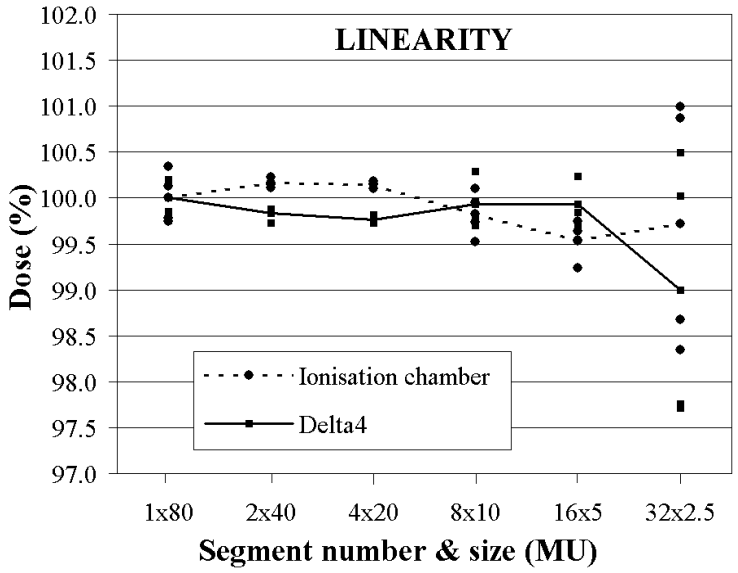

Figure 3

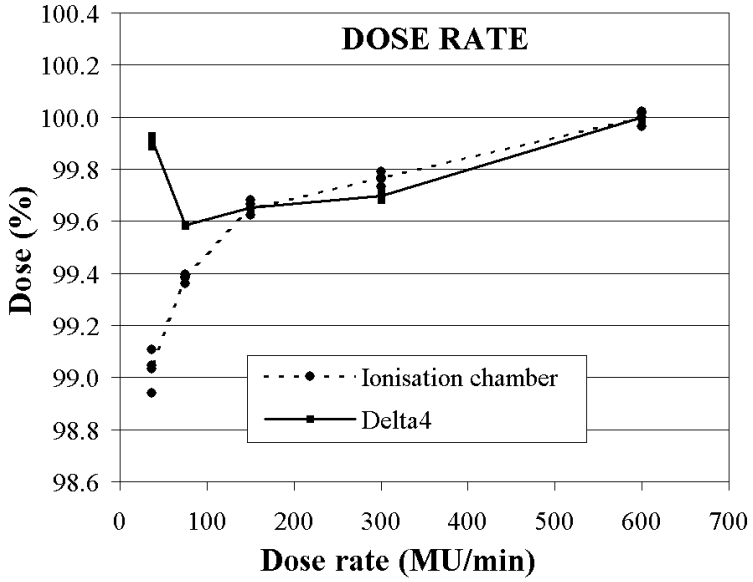

Figure 4

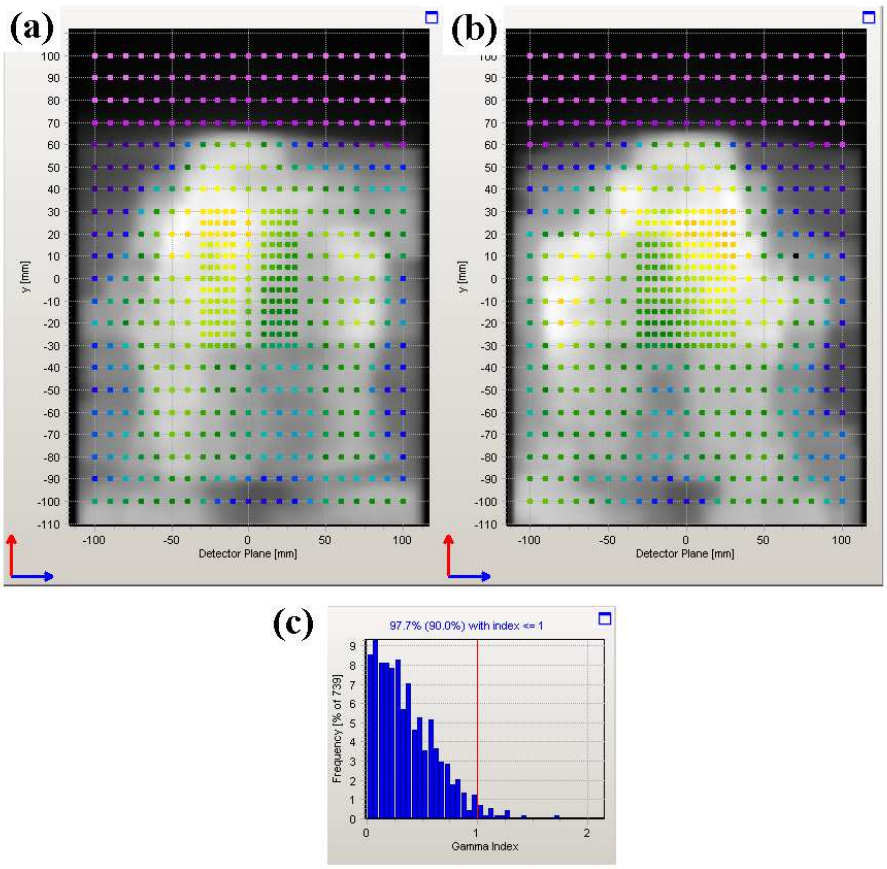
(a)
(b)
y [mm]
Detector Plane [mm]
(c)
97.7% (90.0%) with index <= 1
Frequency [% of 739]
Gamma Index

Figure 5