Multistage evaluation and commissioning of a pre-calibrated, single-use OneDosePlus MOSFET system for *in vivo* dosimetry in a radiotherapy department

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Objectives: The aim was to determine the agreement between absorbed dose measurements using the OneDosePlus and treatment planning system calculations in ideal circumstances, minimising patient-related uncertainties, before deciding upon action levels.

Methods: A OneDose metal oxide semiconductor field-effect transistor (Sicel Technologies, Morrisville, NC) *in vivo* dosimetry system was subjected to a multistage evaluation that tested standard and non-standard field conditions related to treatment planning system calculations for an anthropomorphic phantom.

Results: The system was found to perform within manufacturer specifications. Batch uniformity was found to be within specification when measured using a method described by Halvorsen. A modification used to assess statistical distribution of response showed an increase in the value of two standard deviations (2σ) from ± 2.3 to $\pm 2.9\%$, which was still within the manufacturer-stated value of $\pm 5\%$. The tests using the anthropomorphic phantom also emphasised the fact that patient density inhomogeneities in the region of the $D_{\max}$ point will affect the dose calculated by the treatment planning system and delivered to the patient.

Conclusion: The OneDose system does not account for these inhomogeneities, leading to dependence in the deviation between expected and reported dose on inhomogeneity and choice of calculation algorithm.

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In the 2006 annual report [1], the Chief Medical Officer of the Department of Health recommended that *in vivo* dosimetry (IVD) monitoring should be used at the beginning of each course of radiotherapy treatment; this was later endorsed in the foreword of “Towards Safer Radiotherapy” [2]. IVD has the potential to detect dosimetric errors between the planned and the delivered radiotherapy treatment dose, allowing the remaining treatment fractions to be corrected [1].

Following incidents in recent years, the Scottish Government has adopted this recommendation. In response to this, Ninewells Hospital’s Radiotherapy Department purchased a dosimetry system (OneDose; Sicel Technologies, Morrisville, NC). The OneDose system uses a single-use, pre-calibrated metal oxide semiconductor field-effect transistor (MOSFET) placed on the patient surface on the central axis to measure the maximum incident dose for one beam. OneDose dosimeters have no build-up cap and are designed for use with electrons and photons with added build-up material. OneDosePlus dosimeters (Sicel Technologies) have an integral build-up cap and are used only with photons with no additional build-up material. Previous evaluation tests of the OneDose MOSFET-based IVD system have only considered performance in standard field conditions on solid water phantoms for dosimeters without inherent build-up caps in the megavoltage range for photons. It is

important to note the difference in calibration methods between the dosimeters with and those without build-up caps. OneDose dosimeters are calibrated to report dose to the MOSFET under full electronic equilibrium conditions, which is then source-to-surface distance (SSD) corrected to dose at depth of maximum dose. OneDosePlus dosimeters are calibrated to give dose at the depth of maximum dose directly without application of additional build-up material. The build-up cap, in conjunction with the energy correction factor, is stated by the manufacturer to allow the OneDosePlus dosimeter to be valid for use with photons between 6 MV and 18 MV.

Halvorsen [3] investigated reference condition fields for OneDose dosimeters, but the study did not include an investigation of OneDosePlus dosimeters; in addition, he did not evaluate temperature dependence. Other centres have performed similar evaluations before proceeding to pilot studies on patient treatments to determine action levels (personal correspondence and submitted reports from Scottish centres in response to the Scottish Radiotherapy Advisory Group’s request for current status updates in the action notes of their meeting on 18 June 2009 [4]).

This paper describes the first two stages of a three-stage implementation. The aim was to determine the agreement between absorbed dose measurements using the OneDosePlus and treatment planning system (TPS) calculations in ideal circumstances, minimising patient-related uncertainties, before deciding upon action levels. The first stage builds on the work by Halvorsen [3] and includes an investigation of angular and temperature dependence.

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The second stage considers the effect of non-standard field sizes and geometries, positional dependence in asymmetric fields and the effect of inhomogeneous tissue densities. The effect of the treatment planning dose calculation algorithm on the difference between measured dose and predicted dose is also explored. The third stage will be a pilot study considering practical implementation in radiotherapy. The work presented in this paper comprises the first and second stages.

Methods and materials

The system investigated was a OneDosePlus IVD system. Details of the structure and principles of operation of this MOSFET system are given by Halvorsen [3]. Exposures to 6 MV and 10 MV photon beams were made using linear accelerators (Varian Clinac iX and Varian 600C; Varian Medical Systems, Palo Alto, CA). Field chambers used in photon measurements were 0.60 cc cylindrical Farmer chambers (QADOS, previously NE Technology, Sandhurst, UK). A TPS (Oncentra MasterPlan (OMP) v3.2; Nucletron, Veenendaal, The Netherlands) was used at Stage 2, as was an anthropomorphic phantom (Phantom Laboratory RANDO; The Phantom Laboratory, Salem, NY).

When the system is used, a dosimeter is inserted into a hand-held reader. Beam parameters of radiation modality, energy, dose rate, field size, SSD, monitor units (MU) and wedge angle can be defined via a numeric pad. Definition of these parameters allows the system to apply correction factor equations from the dosimeter's electrically erasable/programmable memory (EEPROM) to the subsequent dose reading. The dosimeters used are MOSFETs with a gate voltage that will allow current to flow through it ("threshold voltage"). The MOSFETs are used unbiased. The dose measured is related to the shift in threshold voltage following exposure to radiation. It is therefore necessary to determine the starting threshold voltage, which is done by zeroing the dosimeter in the reader [3]. All entered information is stored in the dosimeter's EEPROM along with the dose record. In each instance, the dosimeter is zeroed, with the modality, energy, dose rate, field size, SSD and MU defined in the reader before irradiation. Once the dosimeter is read it cannot be re-zeroed, which precludes reuse.

Stage 1

The following tests were performed during Stage 1 of the evaluation to test the system at standard set-ups: dose accuracy; intrabatch uniformity; energy correction validity; angular dependence and temperature dependence.

Dose-rate dependence and dose linearity tests performed by Halvorsen [3], and the evaluation of the build-up effect of the integral build-up cap of the OneDosePlus system by Pursley et al [5] were not repeated in this study. It was assumed that these parameters would be consistent between dosimeter batches.

Dose accuracy

The dose accuracy was tested by comparing the measured dose from OneDosePlus dosimeters and a calibrated Farmer chamber using a 6 MV photon beam.

The Farmer chamber was positioned at a depth of 5 cm in a solid water phantom at 100 cm SSD in a 10×10 cm field and irradiated with 200 MU. Percentage depth dose (PDD) curves for the linear accelerator were used to calculate the absolute dose at $D_{\max}$. Five OneDosePlus dosimeters were then positioned on the surface of the phantom and irradiated sequentially using the same field set-up. SSD, MU, dose rate, field size, modality and energy were all defined via the reader prior to irradiation. The dosimeters were read within 1 min of irradiation and the reported dose compared with the calculated true $D_{\max}$.

Intrabatch uniformity

To measure intrabatch uniformity, 10 OneDosePlus dosimeters were placed 10 cm deep in a solid water phantom on the horizontal plane through the isocentre. The dosimeters were arranged in a 4×1 cm region around the central beam axis. Analysis of beam profiles across this region showed the field flatness $<1\%$ (max-min) [3]. A 1 cm-thick sheet of soft tissue-equivalent bolus material placed directly on top of the dosimeters minimised any air gaps. The dosimeters were irradiated simultaneously with 200 MU by a 20×20 cm field of 6 MV photons at a dose rate of 250 MU min^{-1} . The intrabatch uniformity was measured by calculating two standard deviations (2σ) of the 10 dosimeter readings as a percentage of their mean. This is the parameter compared to that quoted by the manufacturer.

Energy correction

The OneDose system's correction for photon energy was evaluated by comparing dose measured by an ion chamber at depth and corrected by PDD curves to the position of maximum dose, to the OneDose-reported $D_{\max}$ reading for 6 MV and 10 MV fields. The fields used were 10×10 cm, 100 cm SSD with 100 MU. Five OneDosePlus dosimeters were irradiated sequentially with the same field for each energy level. A reference chamber placed in the field in air was used for each irradiation and the air temperature recorded. Readings from this chamber were used to correct for fluctuations in linear accelerator output between set-ups.

Angular dependence

Angular dependence of the system was tested by irradiating OneDosePlus dosimeters positioned at the isocentre in a custom-made cylindrical foam holder at gantry angles of 0° , 45° , 90° and 180° . Two dosimeters were irradiated at each angle.

Temperature dependence

To investigate temperature dependence, two dosimeters were placed on the surface of an in-house temperature phantom and left to reach thermal equilibrium. The phantom comprised a metal box containing an adjustable heating element and a surface contact digital thermometer. The dosimeters were then irradiated *in situ* with a 10×10 cm field, SSD 100 cm, with 6 MV photons for 100 MU with the temperature phantom at room temperature (approximately 23°C). The experiment was repeated at phantom temperatures of approximately 30°C and 40°C .

Statistical distribution of the response of a batch of detectors

Both the manufacturer and Halvorsen [3] use standard deviations and standard error in their analyses; however, no statistical tests have been performed on large samples of OneDose dosimeters (of either variety) to confirm normal distribution of their response. In light of this, 96 OneDosePlus dosimeters were irradiated in 3 batches. Each batch of 32 dosimeters was zeroed and placed in a grid pattern on a 10 cm-thick block of solid water in a 20×20 cm field at 105 cm SDD on the central axis. All dosimeters were at least 3 cm from the field edges. A 1 cm-thick sheet of soft tissue equivalent bolus material placed directly on top of the dosimeters minimised any air gaps and a further 4 cm of solid water was placed on top. A reference chamber was used to correct for linear accelerator output fluctuations between batches and air temperature and pressure readings were recorded. The corrected reported doses were imported into a statistics program and plotted as a histogram, which was analysed to determine normality of distribution.

Stage 2

The following tests were performed during Stage 2 of the evaluation to investigate whether the system remained within manufacturer specifications in non-standard set-ups.

Treatment plans were generated on a CT image set of a RANDO anthropomorphic phantom using the OMP TPS. Four fields were planned. The first 2 fields were a small 5×5 cm field and a large 25×25 cm field. The third field measured 20×12.5 cm with an irregular multileaf collimator (MLC) shaping applied. The fourth field was a 10×10 cm field with a 45° dynamic wedge. All four fields shared a common isocentre and were set at gantry angle of 0° entering the phantom anteriorly (Figure 1). Plans were created for 6 MV and 10 MV photon beams, with dose points placed at the appropriate point of maximum dose on the beam central axis on each plan. The dose contributions from each field were calculated for 200 MU per field. The phantom was positioned on the treatment machine and the fields delivered three times with a OneDosePlus dosimeter placed on the beam central axis (optical crosshair).

The treatment plans were calculated using pencil beam (PB) and collapsed cone (CC) algorithms to provide a broader understanding of variation between reported and expected doses under different clinical situations. The reported doses were compared with the expected dose point doses from the TPS, correcting for the calibration of the machine on the day of measurements (which differed from the ideal of 1 cGy MU^{-1}).

Positional dependence in asymmetric fields was examined by producing plans on the RANDO phantom images using moderately asymmetric and half-beam blocked fields at 6 MV and 10 MV. The field dimensions ($X1/X2/Y1/Y2$) for the two fields were $-5 \text{ cm}/5 \text{ cm}/$

Figure 1. CT slice of RANDO phantom (The Phantom Laboratory, Salem, NY) showing location of planned fields in thorax. (a) Sagittal and (b) coronal views; (c) cross section views of the same.

–4 cm/8 cm for the moderately asymmetric field and 0 cm/10 cm/–6 cm/6 cm for the half-beam blocked field. Dose points were defined at $D_{\max}$ (for the relevant energy level) below the surface of the phantom at the geometric field centre for both fields, and under the beam central propagation axis for the moderately asymmetric field. All fields were positioned over the abdomen to isolate differences due to field asymmetry rather than complications involving tissue inhomogeneity and lateral scatter effects. The phantom was set up in the treatment position and the fields delivered three times with dosimeters in place above the respective TPS dose points.

Results

Stage 1

Dose accuracy

The observed difference between mean dosimeter reading and measured $D_{\max}$ was +3.5%. The manufacturer's system specification states the dose accuracy in the 30–500 cGy range to be $\pm 5\%$ for standard set-up.

Intrabatch uniformity

Reported doses from 10 simultaneously irradiated dosimeters gave 2σ of $\pm 2.3\%$.

Energy correction

Comparison with Farmer chamber measurements gave differences of +3.5% at 6 MV and –2.9% at 10 MV.

Angular dependence

The OneDosePlus dosimeter readings at gantry angles of 45° and 90° were within 1% of those at 0° . At 180° , the dosimeter readings fell by approximately 12%.

Temperature dependence

The temperature dependence data showed no clear linear relationship between dose and surface temperatures ($R^2 < 0.5$). This is consistent with the findings of Best et al [6] for OneDose dosimeters without build-up caps. All results were within 2% of the reading at room temperature (Figure 2).

Statistical distribution

The reference chamber readings corrected to standard temperature and pressure were found to be identical to three decimal places. The reported doses of the three batches of dosimeters did not require correction for output variation.

The final sample size used was $n=94$. Beam profiles at x -axis offsets of –4.3, –0.8, 0.8 and 4.3 cm, corresponding to the lines of the dosimeter placements, showed the field flat to be $<1.7\%$ (max–min). Using the marked positions of the dosimeters on the grid and the zeroing

Figure 2. Plot of relative response of OneDosePlus (Sical Technologies, Morrisville, NC) surface temperature over the clinical useful range. Vertical error bars represent two standard deviations of 2.3% from batch uniformity test. Horizontal bars show precision of the digital thermometer.

Figure 3. Histogram plot of $n=94$ irradiated dosimeter reported doses showing normal distribution with superimposed Gaussian curve. Two standard deviations, $2\sigma=7.1$ cGy (2.9%). OneDosePlus manufactured by Sixel Technologies, Morristown, NC.

time of the dosimeters, correction factors were calculated and applied to the dosimeter results to compensate for any effect of field non-uniformity.

Exploratory data analysis was undertaken via examination of box plots and histograms, and the Kolmogorov–Smirnov test was conducted to assess normality. Statistical significance was set at $p<0.05$.

The outcome of this test was non-significant $D(94)=0.051$, $p>0.05$; thus, normality can be assumed and parametric statistics used (Figure 3).

Stage 2

Non-standard fields

The deviations from expected $D_{\max}$ are given in Tables 1 and 2, for 6 and 10 MV, respectively. Owing to machine time and cost constraints, the MLC field was not carried out for 10 MV.

Positional dependence and asymmetry

Tables 3 and 4 summarise the deviations between reported doses and expected TPS doses, corrected for linear accelerator output, for moderate and half-beam fields. Mean deviations are shown from PB and CC algorithm-expected values. For the moderate asymmetry

field, the mean deviations between geometric centre and central propagation axis were 2.7% for 6 MV and 4% for 10 MV.

Discussion and conclusion

Stage 1

Dose accuracy, intrabatch uniformity and manufacturer programmed energy correction factors were all confirmed to be within manufacturer specifications. The manufacturer advises that the optimal accuracy is attained by reading the dosimeters 2–3 min post-irradiation, but the general specification holds for between 1 and 10 min. Time efficiency is a key consideration in meeting government-recommended targets, so clinical use will most likely dictate reading between 1 and 2 min post-irradiation in order to minimise time delays in patient treatment.

The manufacturer states that OneDose dosimeters can be calibrated and correctable for a range of energies, and the calibrations provided are quoted for nominal or “Machine Nameplate” energy, which is a qualitative (not a quantitative) measure. The energy correction provides a PDD correction to dose at maximum depth. Discrepancy between local and manufacturer definitions of

Table 1. 6 MV beam deviations between reported $D_{\max}$ and expected $D_{\max}$ values, calculated with PB and CC algorithms by TPS

Field description	Field size	Deviation (PB)	Deviation (CC)
Small	5 × 5 cm	−4.9%	−3.5%
Large	25 × 25 cm	−4.4%	−1.7%
Irregular with MLC	12.5 × 20 cm MLC	−4.2%	−2.6%
Dynamic-wedged	10 × 10 cm 45° W	−4.9%	−3.4%

CC, collapsed cone; MLC, multileaf collimator; PB, pencil beam; TPS, treatment planning system; W, wedge.

Table 2. 10 MV beam deviations between reported $D_{\max}$ and expected $D_{\max}$ values, calculated with PB and CC algorithms by TPS

Field description	Field size	Deviation (PB)	Deviation (CC)
Small	5 × 5 cm	−5.3%	−4.1%
Large	25 × 25 cm	−2.0%	0.8%
Dynamic-wedged	10 × 10 cm 45° W	−3.9%	−2.5%

CC, collapsed cone; PB, pencil beam; TPS, treatment planning system; W, wedge.

nominal energy could lead to over or under correction. However, results for 6 MV and 10 MV in this work were found to be within manufacturer specification.

The intended use of the OneDosePlus dosimeters to measure entrance $D_{\max}$ from individual beams means that an incident angle of greater than 90° should not occur. For this purpose, the response of the OneDosePlus dosimeter may be considered to have negligible angular dependence. It should be noted that results reported by Ainsley et al [7] claimed a 60% deviation at a gantry angle of 80°. That work, however, compared dosimeter results to ion chamber readings in a phantom, whereas this work compared OneDosePlus dosimeter responses relative to each other in an isotropic phantom. It is unclear whether the authors moved the ion chamber with gantry angle to maintain a fixed chamber to beam geometry.

Because the 2% range from reading at room temperature is less than 2σ for 23–41°C, the dose accuracy can at least be considered to be maintained between room and skin temperature, which is within this range.

Stage 2

Taking linear accelerator output into consideration, for symmetric non-reference fields, all reported doses were within the 5% of the expected doses except for the small field at 10 MV PB (−5.3%), although 5% of measurements were expected to deviate by more than 5%. This was within the manufacturer specification. Pursley et al [5] showed that the build-up cap provides partial build-up, which is then corrected to dose at maximum depth for the specific energy used. They also reported over and under response for larger and smaller fields at higher energies, suggesting increased electron contamination dose contribution (because the field size correction used was for 6 MV and not the energy level specified by the user).

Analysis of the non-reference field results showed a greater difference between deviations when comparing reported doses with expected doses calculated with PB and CC algorithms for the 25 × 25 cm field than for other fields.

Table 3. Comparison of OneDose (Sicel Technologies, Morristown, NC) vs TPS PB and CC algorithms for a moderately asymmetric field using 6 and 10 MV at locations 1 (central propagation axis) and 2 (geometric field centre)

Energy level	Location	Deviation (PB)	Deviation (CC)
6 MV	1	−2.9%	−1.6%
	2	−1.9%	0.1%
10 MV	1	−6.6%	−5.2%
	2	−4.2%	−2.1%

CC, collapsed cone; PB, pencil beam; TPS, treatment planning system.

The positioning of dosimeters in this experiment (sternum) raised the important question of suitability of citing for dosimeters in clinical conditions. Fields passing through the lungs where the calculation point would be in air would be prone to gross deviations between reported and expected doses.

For the moderate asymmetry plan, the mean deviations of the dosimeter-reported doses for 6 MV were within the $\pm 5\%$ manufacturer specification from the expected dose at both dosimeter positions. At 10 MV, the mean reported doses from dosimeters placed at the geometric field centre were within $\pm 5\%$, while those placed on the propagation central axis deviated by −5.2% from CC algorithm calculation and by −6.6% from PB algorithm calculation.

The full asymmetry plan mean and individual deviations were within manufacturer specifications for all energies and algorithms.

Location 2 appeared to provide marginally better $D_{\max}$ agreement. A 2-tailed Student's *t*-test was conducted and no significant differences were found between the 2 locations at either 6 MV ($t(4)=0.87$, $p>0.05$, non-significant) or 10 MV ($t(4)=1.99$, $p>0.05$, non-significant). Thus, it appears that, for a moderately asymmetric beam, there was no evidence that dosimeter accuracy depends on the choice of location. However, the very small sample size means that this test is under-powered and unlikely to have been able to reliably detect any differences even if they did exist, so these results should be interpreted with caution. All mean and individual deviations from TPS-calculated doses were within the general manufacturer accuracy specification of $\pm 5\%$ for both locations for 6 MV. With this in mind and considering the practicalities of set-up, it was decided that, for moderate asymmetry plans, the central propagation axis should be used in order to enhance reproducibility in dosimeter placement and of the TPS $D_{\max}$ dose point. When considering asymmetric 10 MV beams, it is important to recognise the reduced accuracy of the system and increase any action limits accordingly.

In conclusion, the OneDose IVD system using OneDosePlus dosimeters performs within manufacturer-stated specifications for standard and non-standard field configurations and shows acceptable agreement with

Table 4. Comparison of OneDose (Sicel Technologies, Morristown, NC) vs TPS PB and CC algorithms for a fully asymmetric field at 6 and 10 MV

Energy level	Deviation (PB)	Deviation (CC)
6 MV	0.9%	2.4%
10 MV	−1.8%	0.3%

CC, collapsed cone; PB, pencil beam; TPS, treatment planning system.

expected data from TPS calculation on a static anthropomorphic phantom. The internal density configuration in the region around the $D_{\max}$ point and the calculation algorithm used must be considered when interpreting reported doses from the system relative to planned dose.

OneDosePlus dosimeter response was found to be normally distributed, allowing any future studies to utilise parametric statistics. Batch uniformity measured for $n=94$ dosimeters in a larger field region showed 2σ to be 2.9%.

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