

Evaluation of precalibrated implantable MOSFET radiation dosimeters for megavoltage photon beams

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We have studied the response of factory calibrated implantable MOSFET detectors to absorbed doses from 100 to 400 cGy. The average measured dose is quite close to the true delivered dose, with the standard deviation falling between 1.4 and 3.6%. The measured dose tends to be slightly underestimated for smaller doses, while it tends to be slightly overestimated for larger doses. Thus, although the calibration of the detector is most accurate for doses close to the calibration dose of 200 cGy, it may be used over the range of commonly used doses in fractionated radiotherapy. © 2005 American Association of Physicists in Medicine. [DOI: 10.1118/1.2065447]

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I. INTRODUCTION

The utility of a radiation detector or dosimeter depends strongly on its robustness over a range of conditions. For absorbed dose measurements, a detector is only as good as the accuracy of its calibration. Further, although a detector may be individually calibrated for quality assurance applications, one that is precalibrated and ready for immediate use would be ideal for clinical applications.

Recently, a new metal-oxide-semiconductor field-effect transistor (MOSFET) dosimeter has been developed for use during external beam radiation therapy.^{1,2} The dosimeter is small (3 mm diameter, 25 mm length), wireless, and designed to be permanently implanted *in vivo*. Unlike the commonly used detectors and dosimeters to date such as thermoluminescent detectors, photodiodes, and most other MOSFET detectors, it is factory calibrated. We have confirmed the reproducibility of the detector response following individual calibration,³ in which the cumulative changes in threshold voltage for each detector was fit to the cumulative dose to obtain an equation for the dose per treatment fraction. Here we investigate the validity (accuracy, precision, and applicability) of the factory calibration of these detectors for a range of absorbed doses from 100 to 400 cGy.

II. METHODS AND MATERIALS

A. Description of the detector and its factory calibration

The detectors used in this study are prototypes manufactured by Sixel Technologies (Morrisville, NC). The radiosensitive area of the detector is a *p*-channel MOSFET with a 4000 Å oxide layer. The MOSFET, as well as a data acquisition chip, a microprocessor, and a copper coil, is encapsulated in a glass tube 3.25 mm in diameter and 25 mm in length. The circuit is powered by a current induced in the coil by an external handheld antenna connected to an rf reader. The dosimeter is passive during irradiation and powered only during measurement of the threshold voltage of the

MOSFET. The microprocessor controls both data acquisition and reader/dosimeter communication. A laptop computer controls the rf reader and converts the digital signal to a decimal voltage. The setup is depicted in Fig. 1.

During factory calibration,¹ a number of detectors are taken from a particular manufacturing lot and exposed to a series of 200 cGy irradiation fractions under 6 MV x rays. In order to mimic realistic *in vivo* treatment conditions, the detectors are held at constant temperature throughout the time of calibration, with at least a one-day interval between fractions. Because the absorbed dose is proportional to the change in threshold voltage due to irradiation, the threshold voltage data is fit to the cumulative absorbed dose. This fit accounts not only for the nonlinearity in response with increasing dose^{4,5} but also for the fading of the threshold voltage^{3,6,7} between irradiation fractions. There is also a small correction of less than 1% to account for the anisotropy of the detector with respect to the relative positions of the beam direction and the long axis of the dosimeter.¹ The fit is used for all dosimeters in the manufacturing lot, with small corrections for individual detector response determined following a 1000 cGy predose.¹ In order to determine the measured dose for each irradiation fraction, measurement of the pre- and post-irradiation threshold voltages is required; no further information (e.g., cumulative dose, time elapsed between irradiation fractions) is needed.

B. Experiment

The MOSFET dosimeters, all from the same manufacturing lot, were provided by Sixel Technologies, Inc. Two detectors at a time were placed in an acrylic phantom⁸ (Fig. 2) (Quasar body phantom, Modus Medical Devices, London, Ontario, Canada) at depths to obtain the desired doses of 100, 200, or 400 cGy. The detectors were held at a fixed orientation and a constant temperature of 37 °C (body temperature) throughout the experiment. All measurements were performed using a Theratron 780-C cobalt machine (Atomic Energy of Canada Limited, Mississauga, Ontario, Canada) at

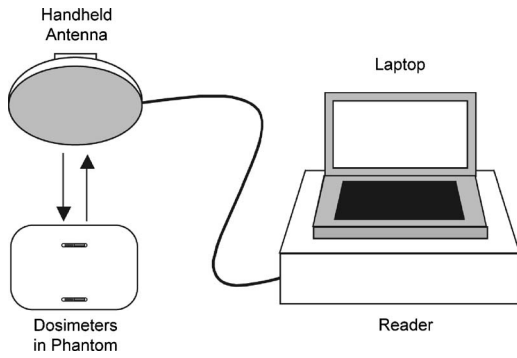


FIG. 1. Schematic of the radiation detection system.

the standard source-to-surface distance (SSD) of 80 cm and a radiation field size of $10 \times 10 \text{ cm}^2$. Each detector was irradiated for 20 fractions, with at least a one-day interval between fractions. Immediately before irradiation, and one to two minutes following irradiation, the detectors were biased for reading with an external radiofrequency reader, and the resulting threshold voltages were converted to absorbed dose using the manufacturer’s calibration. We have also assessed the degree of fading in the threshold voltage for the two detectors exposed to 200 cGy per treatment fraction by measuring the threshold voltage at various intervals over a 24-h time period following irradiation.

III. RESULTS AND DISCUSSION

A. Fading of the threshold voltage following irradiation

Irradiation of MOSFETs causes the creation of electron-hole pairs in the transistor’s oxide layer, and it is the trapped

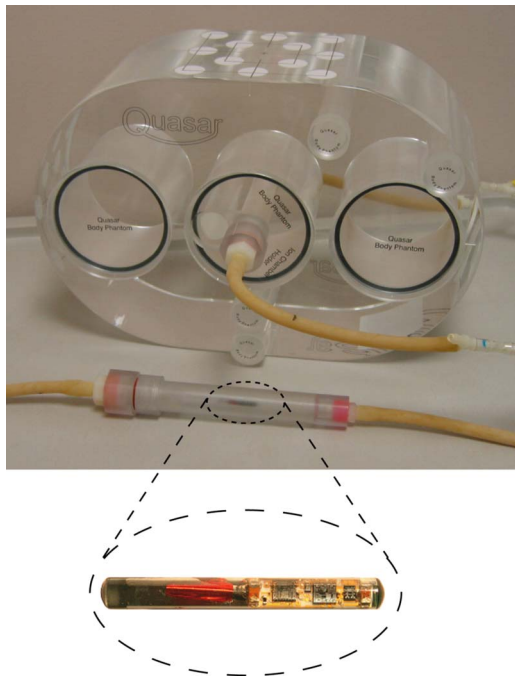


FIG. 2. Acrylic phantom and implantable detectors. Each implantable detector is placed in a tube (shown in the foreground) and surrounded by a constant flow of water to maintain body temperature. An enlarged view of the detector (courtesy of Sixel Technologies) is also shown.

TABLE I. Degree of fading for the first dosimeter irradiated at 200 cGy/Fx. Change in threshold voltage ($\Delta V_{T,\text{Fading}}$) is given relative to the reading within 2 mins following irradiation. The average response of the detectors over 20 fractions was 0.39 mV/cGy.

Elapsed time	$\Delta V_{T,\text{Fading}}$ (mV) Fraction 1	$\Delta V_{T,\text{Fading}}$ (mV) Fraction 15
5 min	1	1
10 min	1	3
6 h	7	17
24 h	12	26

holes that cause the increase in the absolute value of the threshold voltage. Fading of the threshold voltage occurs due to recombination of the electron-hole pairs and has a natural logarithmic dependence on time.^{9,10} We have also observed this logarithmic behavior with the implantable detectors. Table I shows the degree of fading in the threshold voltage 5 min, 10 min, 6 h, and 24 h following delivery of the first and 15th irradiation fractions for one of the implantable detectors irradiated to 200 cGy per fraction. We can see from the table that the threshold voltage changes little over the first ten minutes, and after 6 h the threshold voltage decays up to 65% of the value after 24 h. After the 15th irradiation fraction, the degree of fading is more than twice that following the first irradiation fraction. This may occur due to the increase in trapped holes with accumulated dose, leading to a greater probability for electron-hole recombination.

B. Evaluation of the calibration at 200 cGy

In order to test the validity of the factory calibration at the calibration dose of 200 cGy, we have exposed two detectors to 200 cGy per treatment fraction for 20 fractions. The results are shown in Fig. 3. Table II provides a summary of results for all detectors. The average daily readings for the two detectors were 200.7 and 199.4 cGy, with respective standard deviations of 1.4% and 2.1%. Although this error is slightly higher than for detectors whose response was determined from individual calibrations,³ the factory calibration appears to work quite well in this dose range. These results

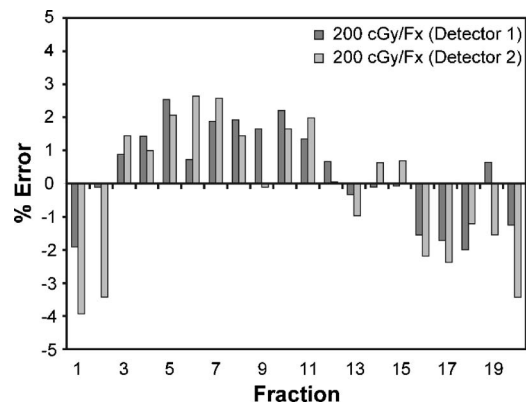


FIG. 3. Percent error in the dose measured by two implantable detectors using the manufacturer’s calibration for a delivered dose of 200 cGy.

TABLE II. Summary of results for all detectors over 20 irradiation fractions.

	Detector 1	Detector 2	Detector 3	Detector 4
Dose/Fx(cGy/Fx)	200	200	400	100
Average reading (cGy)	200.7	199.4	403.7	96.7
Std. Dev. (%)	1.4	2.1	2.3	3.6

compare well with the manufacturer's stated standard deviation of 3.5%, determined for similar doses per fraction.^{1,2} Aside from set-up errors, a maximum error of 1% could be expected due to the round-off error in the analog-to-digital converter in the data acquisition chip of the detector. Errors of less than 1% arise from the uncertainty of the readings from the temperature controller (0.1 °C), and the temperature dependence of the threshold voltage of the MOSFET detector (4.3 mV/°C).

C. Validity of the calibration for larger and smaller doses

Here we have examined the validity of the factory calibration for doses significantly larger and smaller than the dose at which these detectors are calibrated (200 cGy). For this purpose, we have selected the doses of 400 and 100 cGy per irradiation fraction. The results are summarized in Table II. The larger dose per fraction of 400 cGy results in a slight overestimation in dose measurement (Fig. 4), with an average daily reading of 403.7 cGy, and a standard deviation of 2.3%. Using a smaller dose of 100 cGy leads to an underestimation in dose measurement, in which the average daily reading was 96.7 cGy and the standard deviation was 3.6%. The main cause of the deviations is the nonlinear response of the detector with accumulated dose,^{1,3} in which the change in threshold voltage decreases with increasing accumulated dose. This is a general property of MOSFETs.^{4,5} The calibration function describing the measured dose depends not only on the dose per treatment fraction, but also the fading between fractions. Since fading occurs due to electron-hole recombination, it is itself dependent on the number of trapped holes and therefore on both the absorbed dose for a given

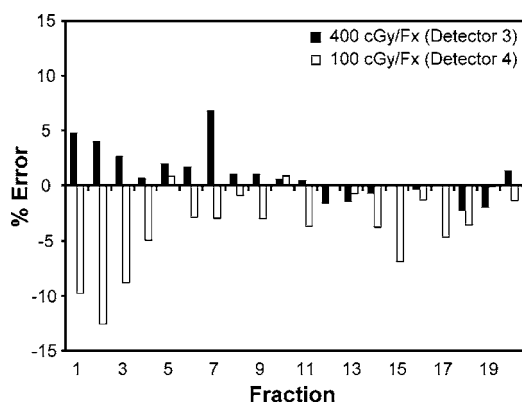


FIG. 4. Percent error in the dose measured by two implantable detectors using the manufacturer's calibration for respective delivered doses of 100 and 400 cGy.

irradiation fraction and the total accumulated dose. It is therefore reasonable that some deviation in measured dose would occur for absorbed doses that are significantly larger or smaller than the calibration value. The larger standard deviation of the detector irradiated to 100 cGy can be attributed to the magnified effects of round-off error and temperature on the threshold voltage. The highest error occurs in the first three irradiation fractions. This may be due to the significant amount of elapsed time (days or months) between the final factory irradiation fraction and the first clinical irradiation fraction, allowing for significant fading in the threshold voltage and resulting in smaller than expected values of the threshold voltage for the first irradiation fraction. The calibration function (a third-order polynomial) produces a smaller response for lower values of the threshold voltage, which may have led to the underestimation in measured dose for the early irradiation fractions. However, the change in threshold voltage due to fading following the first irradiation fraction was small, leading to a closer match of the response of the detector with the calibration function for subsequent fractions. Looking at the results for all the detectors, it is apparent that there is a small effect on the dose response for conditions that deviate from those used for the factory calibration. However, we have found the calibration to be adequately robust for clinical applications over the range of doses studied here.

The manufacturer's calibration takes into account the change in threshold voltage due to both irradiation and fading, so that the calibration can be applied to an entire lot of detectors to be used for fractionated radiotherapy dosimetry. With the present calibration technique, the only required data are the pre- and post-irradiation threshold voltage readings for a particular fraction. Although more sophisticated techniques may improve the performance of the detector over a larger range of delivered doses by, for instance, including the effects of fading more accurately, such techniques would require more data, including time elapsed between treatment fractions, cumulative dose, etc. On a practical level, what would happen if the user neglected to take a reading for one day, two days, or a week? Considered in this way, more accurate calibration techniques may be too impractical for clinical use. However, for applications where the doses are much larger than those encountered in external beam radiotherapy, such as those prescribed in intraoperative radiotherapy, corrections to the factory calibration must be carefully considered.¹¹

D. Discussion: Comparison with the OneDose™ sensor

There seems to be greater dose dependence on the implantable detector than with the OneDose™ sensor,¹² which is also manufactured by Sixel Technologies. The OneDose, a disposable detector designed for skin dose measurements, also uses a MOSFET to measure the dose from one single irradiation fraction of up to 500 cGy. The higher accuracy of the single-use detector occurs because it is calibrated at the zero temperature coefficient (ZTC) bias current, where the

MOSFET becomes temperature independent.¹³ Unfortunately, the ZTC bias current of the MOSFET in the implantable detector is too high to allow adequate range for the multifraction dosimetry that is required of the detector. Furthermore, the ZTC bias current would vary significantly with the high accumulated doses expected for the implantable detector. Improvements in the design of the detector¹ (including replacing the MOSFET with a dual-transistor design with less temperature sensitivity, adding an on-board temperature sensor to correct for pre- and post-irradiation temperature fluctuations, increasing the accuracy of the analog-to-digital converter, and reducing the overall size of the detector) should significantly reduce the errors observed in the present study.

IV. CONCLUSION

We have studied the factory calibration of implantable MOSFET dosimeters and found the error in the measured dose to be reasonable for clinical applications. The factory calibration of the dosimeters is most accurate for absorbed doses in the range of the dose used for calibration. On average, small decreases in the measured dose may be expected for absorbed doses smaller than that used for calibration, while small increases may be expected for greater absorbed doses. It will be interesting to see how changes in the second-generation devices decrease the range of error of these detectors.

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