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Improved spatial resolution by MOSFET dosimetry of an x-ray microbeam

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Measurement of the lateral profile of the dose distribution across a narrow x-ray microbeam requires a dosimeter with a micron resolution. We investigated the use of a MOSFET dosimeter in an "edge-on" orientation with the gate insulating oxide layer parallel to the direction of the beam. We compared results using this technique to Gafchromic film measurements of a 200 micrometer wide planar x-ray microbeam. The microbeam was obtained by using a vernier micrometer-driven miniature collimator attached to a Therapax DXT300 x-ray machine operated at 100 kV_p. The "edge-on" application allows utilization of the ultra thin sensitive volume of the MOSFET detector. Spatial resolution of both the MOSFET and Gafchromic film dosimeters appeared to be of about 1 micrometer. The MOSFET dosimeter appeared to provide more uniform dose profiles with the advantage of on-line measurements. © 2000 American Institute of Physics.

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I. INTRODUCTION

A major advantage of the metal-oxide-semiconductor field effect transistor (MOSFET) as a radiation monitor is that the radiation-sensitive region, the oxide film, is very small.^{1,2} The sensing volume is much smaller than competing integral dose measuring devices such as the ionization chamber, semiconductor diode or thermoluminescent dosimeter (TLD). The smallest available liquid ionization chamber³ has a dosimetric volume of about 2 mm³, while TLD volume is about 1 mm³. The semiconductor diode sensitive volume is about 0.3 mm³ (Ref. 3). The MOSFET's sensitive volume is typically 1×200×200 micrometers or only 4×10⁻⁵ mm³. Attention is thus being turned to the use of MOSFETs especially where the sensor has to be inserted into a confined space, such as a catheter.^{4,5} This property of the MOSFET also makes it attractive for measurements in the high gradient radiation field where the gradient mostly depends on a single space coordinate, like resolving dose profiles of x-ray microbeams or depth dose distribution.

Using synchrotron radiation, Slatkin *et al.* at Brookhaven National Laboratory⁶ have established that induced brain tumors in rats can be controlled by microbeam radiation

therapy (MRT).^{7,8} The principal behind MRT is that normal tissue can tolerate high doses of radiation without leading to necrosis. This is because the survival of capillary cells adjacent to the exposed region of lethally irradiated capillaries allows regeneration avoiding tissue necrosis.⁹ The high dose delivered in each fraction is sufficient for fast killing of the cells in the path of the microbeam. The exact mechanism of selective tumor suppression by a microbeam is not yet known. At this stage we can only speculate that the difference in growth kinetics between tumor and endothelial cells allows the capillaries to regrow between the dose fractions.

An experimental measurement of absorbed dose distribution across the path of a microbeam represents a challenge as it requires a dosimeter with micron resolution. For this reason, microbeam dosimetry research has focused on Monte Carlo simulations of the absorbed dose.^{6,10,11} The EGS-4 Monte Carlo code¹² was used in most of the simulations. A Monte Carlo simulation relies on a particular energy spectrum of the radiation and some other assumptions. There is a need for experimental verification of lateral dose or intensity distributions. One experimental approach was to measure the radiation dose across a 150 μm planar microbeam with a 5 cm diameter NaI(Tl) scintillation detector.¹⁰ The integral

photon signal was measured versus the micron step displacement of a lead shutter. While providing data on the width of the microbeam, this method lacked the resolution to measure the absorbed dose distribution across the microbeam. The integral method did not allow microbeam penumbra measurements.

The work reported here utilizes the very small sensing volume of the MOSFET to best effect as well as enabling a comparison with Gafchromic film as part of the development and testing of high resolution dosimetry of planar x-ray microbeams.

II. MATERIALS AND METHODS

Three separate dosimeters were selected for the experiments.

- (1) *n*-channel metal oxide semiconductor field effect transistor (MOSFET) dosimeters produced by Detector Ltd., Ukraine with thick gate oxide.
- (2) *p*-channel MOSFET type TOT500 with thick oxide, called RADFET manufactured by REM Oxford, UK.
- (3) Gafchromic film MD-55-1 (Nuclear Associates, Carle Place, NY).

A. MOSFET detectors

The sensitive element of a MOSFET detector is a silicon oxide layer underneath the aluminum transistor gate.^{13,14} The geometry of the sensing regions of the silicon dioxide is slightly different for the two MOSFET devices used. In the case of the TOT500 *p*-MOSFET device, the gate oxide has a serpentine shape, i.e., the source and drain regions have the form of interlocking fingers separated by the gate oxide ribbon layer, packed into a $180 \times 270 \mu\text{m}^2$ rectangular shape. The gate on the *n*-MOSFET device has a rectangular ribbon shape with outer boundary dimensions of about $200 \times 200 \mu\text{m}^2$. The gate thickness is about $1 \mu\text{m}$ for both detectors.

The electrical signal used as the dosimetric parameter of MOSFET detectors is the “threshold voltage.” This parameter exhibits a shift when the device is irradiated. Dose is usually determined by looking up tables or the calibration curve of threshold shift versus dose for the MOSFET lot used. In our experiment the radiation exposures were made in the active mode (with a dc field across the oxide) to increase the sensitivity and linearity of the MOSFET dosimetric characteristics.¹⁵ The gate bias used in these experiments was +5 V for all MOSFETs. This corresponds to a sensitivity of about 5 mV/cGy(TE) for measurements at a depth of 1.5 cm in the solid water phantom in a 6 MV photon beam from a medical LINAC. The voltage shift is measured for a fixed point on the MOSFET current–voltage (*I*–*V*) curve which is chosen to minimize the temperature sensitivity of the readings. In these experiments a pulsed current readout system was used for the *n*-MOSFETs and a nonpulsed current system as recommended by the manufacturer was used for the *p*-MOSFET (RADFET). All measurements were normalized to the maximum response for each particular set of data and absolute dosimetry was not performed. A nonlinearity correction of the MOSFET for each particular set of beam

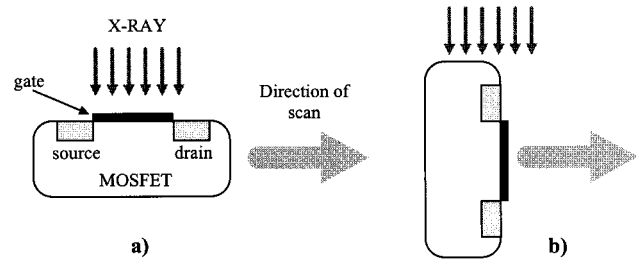


Fig. 1. MOSFET application in (a) “normal” and (b) “edge-on” orientations.

profile measurements was not essential due to the relatively small change of threshold voltage although the correction curve for the single MOSFET detector is well characterized.¹⁶ The microbeam profile graphs were not noticeably changed when the correction was taken into account. The *n*-channel MOSFET chip was mounted inside a Kovar package with the lid removed as described in Rosenfeld *et al.*² The RADFET has a different topology than the *n*-channel MOSFET and was mounted on a plastic board under an epoxy bubble.

The MOSFET measurements were performed in air or with the MOSFET mounted inside the perspex phantom. For in air measurements the MOSFET was mounted at the end of a perspex rod with the sensitive element directly exposed to the x-ray without any build-up material. Since moving a solid detector with a micron step inside a solid phantom is not possible a special micrometer jig was constructed. A MOSFET chip or the phantom as whole was attached to the jig and moved in increments of $10 \mu\text{m}$ to $100 \mu\text{m}$ across the width of the microbeam. The threshold voltage was measured immediately after each irradiation in order to minimize drift effects.

A MOSFET dosimeter is typically used with the surface of the silicon chip normal to an incident x-ray beam (Fig. 1); we will call this a “normal oxide film mode.” For this mode the spatial resolution is limited by the dimensions of the channel—which is, effectively, defined by a rectangle enclosing the source and drain junctions. Our new approach was to rotate the detector plane until the oxide film was “edge-on” to the beam. In this orientation, the theoretical limit of resolution is about $1 \mu\text{m}$, the thickness of the gate oxide.

B. Gafchromic film

Gafchromic film is effectively grainless with a radiographic image spatial resolution of 1200 lines/mm.¹⁷ In our experiment the radiation induced change in light transmission through the film was digitized and analyzed by a two dimensional CCD image analysis system originally designed for astronomical photographic plate image analysis. The $8 \mu\text{m}$ spatial resolution of the readout system exceeded the best currently available densitometer resolution of $20 \mu\text{m}$.¹⁸ The light transmission through the film was measured for filtered red light. The cyan–magenta–yellow (CMY) color filtered light source was set to C=0, M=151, and Y=151.

The median wavelength of the filtered light, as measured by a Jarrel–Ash diffraction spectrometer was 620 nm and the bandwidth was ± 35 nm. In a comparative study of a Gafchromic film dose response¹⁹ for a He–Ne laser densitometer (wavelength 632 nm), a filtered red light densitometer and a broadband densitometer, the greatest response was demonstrated for filtered red light. For calibration of the dose response of Gafchromic film, the 1 cm² films were uniformly irradiated in air to doses from 0 to 40 Gy in steps of 5 Gy. For in air measurement the film strip was suspended between two foam blocks to minimized backscattered radiation. After storing the films at room temperature in a light tight envelope for two days, the relative intensity of the transmitted light through the film was digitized and analyzed by the CCD image analysis system. The dose (D) was plotted versus $-\log(I/I_0)$ where I is intensity of transmitted light through an exposed film and I_0 is intensity of transmitted light through the unexposed film. The calibration results were best fitted by a second degree polynomial.

C. Radiation source

The microbeam chosen for this work was a planar x-ray beam because it was the closest representation of the synchrotron beam that is intended for use in MRT at BNL. A Therapax DXT300 orthovoltage x-ray machine, equipped with a variable slit collimator, was used for generation of this beam. A Therapax DXT300 contains an internal x-ray dosimeter (PTW Diamentor pancake chamber) located between the filter and collimator. The electron beam produced in the Therapax DXT300 hits a target at an angle of 45°. The accelerator was used in service mode at 100 kV_p with a 10×10 cm² collimator and 0.31 cm Al filter. The effective energy of the beam was determined by measuring the aluminum and copper half value layers which were 3.5 mm and 0.15 mm, respectively. Using the energy dependent interaction coefficients for aluminum and copper, as tabulated by Johns and Cunningham,²⁰ together with the measured half value layers yields an effective energy of 39 keV. The size of the electron focal spot on the target is nominally 0.5 cm×0.5 cm according to manufacturer specifications. A variable width collimator with vernier adjustment was fitted to the Therapax collimator. The collimator was set to 200 μm. The length of the slit was 1 cm, to produce a 200 μm planar microbeam. A sketch of the x-ray beam geometry of the Therapax DXT300 orthovoltage accelerator is shown in Fig. 2. The diameter of the anode spot was 0.5 cm and the distance from the spot to the collimator was 50 cm. This geometry was used for beam profile calculation at different detector-to-collimator distances (d) for comparison with experimental data. The calculation program took into account the geometry of the experiment and not the physics of x-ray absorption and scattering. The Therapax x-ray machine was calibrated so that 1 monitor unit (MU) is equivalent to 1 cGy of dose to tissue equivalent material at a point 30 cm from the collimator when a 5 cm diameter conical collimator is used. With the microbeam collimator attached to the Therapax the dose per MU is reduced. The irradiation time of the MOSFET dosimeter for

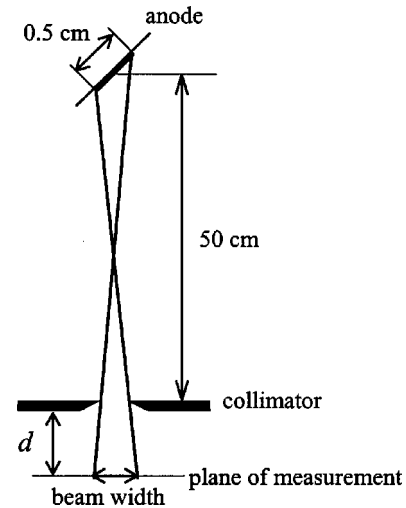


Fig. 2. X-ray beam geometry of the Therapax DXT300 orthovoltage unit.

each data point was 15 to 20 seconds at the x-ray machine dose rate of 0.5 to 1 MU per second depending on the beam current.

The accuracy of microbeam measurements consists of the accuracy of the detector, accuracy of the threshold voltage readout device, accuracy of the positioning device and accuracy of dose delivery by the Therapax x-ray machine. The consistency of dose delivered for each data point was insured by a set beam current on the x-ray machine and a constant irradiation time. This was checked by the internal ionization chamber. The accuracy of the MOSFET threshold voltage readout system was ± 2 mV for a 150 mV threshold voltage change. This is an uncertainty of 1.5%. It was reported elsewhere (see, for example, Kron *et al.*¹⁶) that reproducibility of MOSFET measurements is not worse than the accuracy of the threshold voltage measurement. During this study we irradiated the MOSFETs twice with a fully open collimator to a reference dose of 15 MU. The duration of radiation exposure was 15 seconds. In both cases the change in threshold voltage was 117 mV. The accuracy of the micrometer positioning device was ± 1 μm for movement in the forward direction. For return movement screw backlash reduced the accuracy. Microbeam measurements were performed in the forward direction only. All the components of the system accuracy, including the backlash, were tested by moving the MOSFET 400 μm back to its starting position and repeating the measurement. The accuracy was found to be 3%.

III. RESULTS AND DISCUSSION

The *n*-MOSFET dosimeters were exposed to the microbeam in air, i.e., without any buildup material, in “normal” and “edge-on” orientations. The *p*-MOSFET dosimeter was exposed in the “edge-on” orientation only. The dose delivered to the MOSFET per single irradiation shot in free air geometry was 17 MU which corresponds to a maximum threshold voltage shift in the center of the beam of about 150 mV. The readings were normalized on the maximum MOSFET response for each dose profile. The results (Fig. 3)

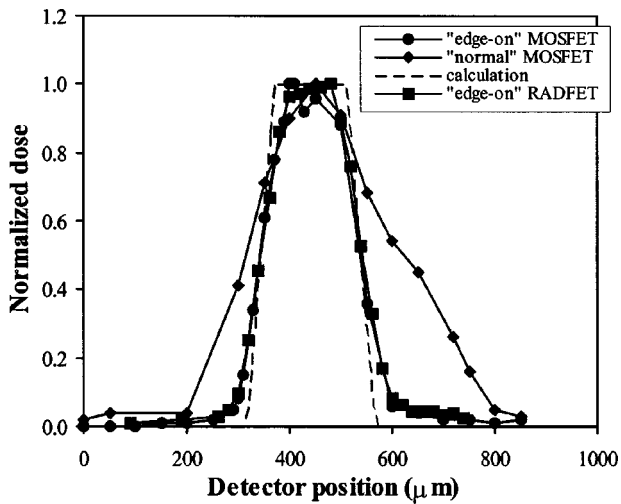


FIG. 3. A comparison of the transverse radiation dose profile across the 200 μm wide microbeam measured by the MOSFET detector in both "normal" and "edge-on" orientations.

prove that the spatial resolution of a MOSFET dosimeter is indeed affected by the orientation of the detector in the beam, the resolution being superior for the "edge-on" mode. The *p*- and *n*-MOSFET curves closely follow each other being nearly identical. This demonstrates that scattering from the MOSFET housing has a little effect to the device resolution for low energy x-rays.

The measurements of a microbeam in a perspex (PMMA) phantom using the "edge-on" MOSFET method were performed at depths in PMMA of 0.8, 3, and 5 cm. Results at the depths of 0.8 and 3 cm are shown in Fig. 4, together with the calculated profiles at the same distance from the collimator. The transverse dose profile at 0.8 cm depth follows the calculated geometrical profile well (Fig. 4). The full width at half maximum (FWHM) of the calculated and measured beams at the depth of 0.8 cm in the phantom are close (Fig. 4). The experimental FWHM exceeds the geometrical one by less than 5%. There is some discrepancy in the beam penumbra because of the partial penetration of x-rays through

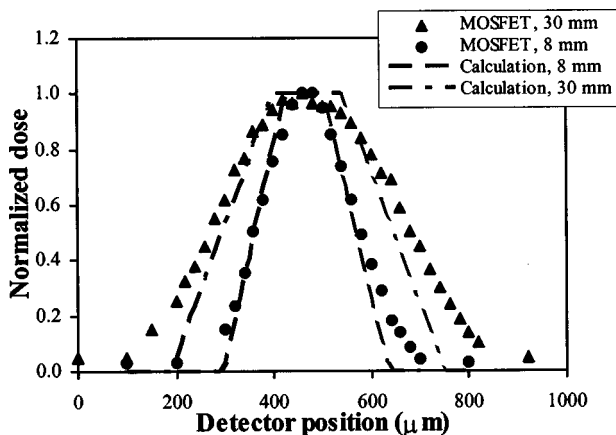


FIG. 4. Measurement and calculation of a transverse microbeam profile at 0.8 and 3 cm depth in the perspex phantom. Measurement by an *n*-MOSFET dosimeter in the "edge-on" mode.

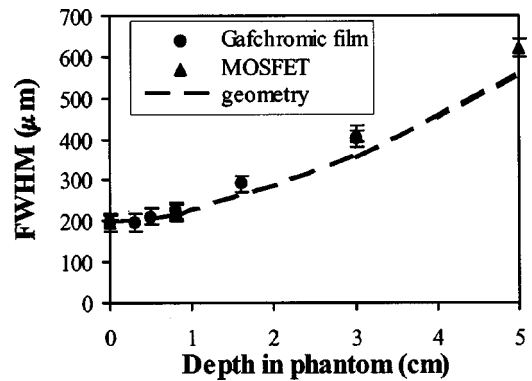


FIG. 5. Experimentally measured and calculated FWHM of 200 μm microbeam in perspex phantom, using Gafchromic film and MOSFET dosimeter in the "edge-on" mode.

the wedge like shape of the variable width collimator plates (Fig. 2). The good agreement between the calculation and measurement indicates that at shallow depths the beam spreads out in the phantom, mostly because of the geometric divergence of the beam. However, the scattering effect becomes more significant at the greater depth. For the depth of 3 cm the experimental FWHM becomes 16% wider than the calculated FWHM.

A beam with low or even no divergence is important for MRT since overlapping of microbeams at some depth in tissue must be avoided. The beam divergence is proportional to the distance between the detector and collimator and inversely proportional to the distance between the beam source and collimator. For the Therapax DXT300 the latter distance is only 50 cm. So far the radiobiological aspects of MRT have been studied at the synchrotron at BNL.^{8,21} For the BNL synchrotron the beam travels 30 m after leaving the synchrotron wiggler and before entering the treatment room. An irradiation specimen at the BNL MRT facility is placed close to the therapy multislit collimator²¹ thus minimizing the divergence.

The measured and calculated FWHMs of the 200 μm planar x-ray microbeam are plotted in Fig. 5 against the depth in the phantom. For the depths of less than 1 cm the increase in microbeam field size due to in-phantom scattering is negligible. Perspex has a higher density than tissue, $\rho = 1.18 \text{ g/cm}^3$, and this has to be taken into account in estimating the scattering effect in tissue. Even considering the 18% difference in density, scattering is not a significant factor in assessing a microbeam width increase for depths in tissue of less than 2 cm.

Both MOSFET and Gafchromic film data are shown in Fig. 5. The dependence of MOSFET and Gafchromic film response on the energy of x-rays should be noted, with MOSFETs demonstrating a stronger dependence.¹⁶ We did not correct for this energy dependence because the FWHM was determined from relative dose measurements at the same depth in the phantom. We assumed that the x-ray spectrum in a certain depth did not change with less than 0.1 cm lateral displacement.

Microbeam measurements by MOSFET and Gafchromic

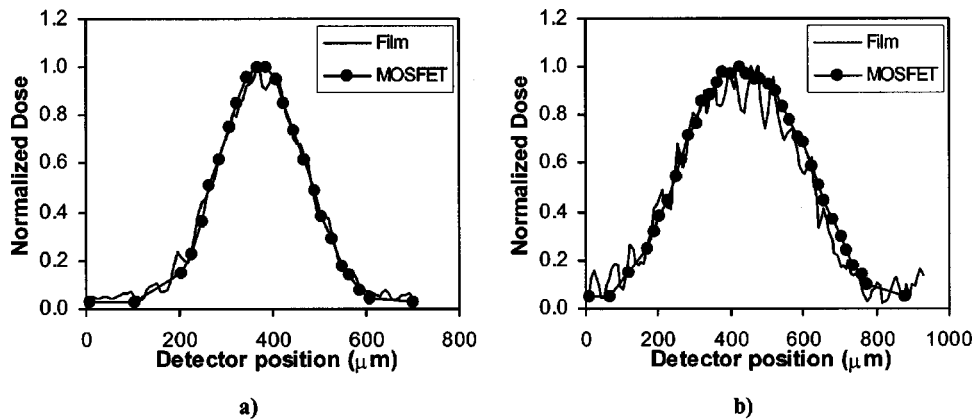


FIG. 6. A comparison of the transverse microbeam profiles as measured by the MOSFET and Gafchromic film at (a) 0.8 cm and (b) 3 cm depth in perspex phantom.

film have very similar FWHM. To compare the two detectors, we measured the transverse profiles at the same depths of 0.8 and 3 cm in the phantom (Fig. 6). In both cases, the FWHM was similar and for the smaller depth [Fig. 6(a)] both the profiles closely follow each other. However, the Gafchromic film data are less uniform. For a greater depth and as a result a weaker signal [Fig. 6(b)], the nonuniformity in the Gafchromic film response becomes more significant, being as high as 20%. A nonuniformity of the optical density of up to 15% of irradiated Gafchromic film was previously reported.^{22,23} Those measurements were performed on a scale of several millimeters, whereas our measurements are on a smaller scale. Narrow spikes in optical density were observed by Meigooni *et al.*²² In this study we were able to resolve these spikes with the distance between neighboring maxima of 40 to 60 μm .

A dose profile as measured by a radiation detector is a convolution of the true dose profile, detector resolution and the readout system resolution. The “edge-on” MOSFET and Gafchromic film dose profiles were identical (Fig. 6). MOSFET was shifted across the microbeam with a step of 10 μm which is similar to the film optical readout system resolution of 8 μm . The resolution of the film is about 1 μm . This implies that the MOSFET resolution limit is also about 1 μm . This is consistent with the main physical limitation of MOSFET resolution which is the 1 μm thickness of the dosimetric volume, the gate oxide. MOSFET resolution can be further improved by using a MOSFET with a 0.1 μm thick oxide layer and by using a scanning system with a sub micron step. The possibility of further reduction of Gafchromic film spatial resolution is limited.

The scanning of a MOSFET point detector takes longer than competing measurement techniques, because a separate short irradiation is required for each data point. However, we have demonstrated that the entire measurement, automated data analysis and curve plotting can be completed in one hour. Gafchromic film analysis took a longer time.

The presence of packaging material around the sensing chip did not seriously affect the spatial resolution at low x-ray energy. Furthermore, in devices designed for “edge-on” scanning, any such effects can be still reduced further, e.g., by encapsulating in very thin materials, reduction of

substrate thickness with machining of the chip to optimize silicon dioxide film exposure, etc.

IV. CONCLUSIONS

We have demonstrated that a radiation-sensing MOSFET detector in the “edge-on” configuration is very suitable for scanning the dose profile of a planar x-ray microbeam. The main advantage over conventional dose integrating methods (TLD, film) arises from the ultra-small size of the sensitive element of the MOSFET structure in the “edge-on” configuration. Another convenience is the complete compatibility of the MOSFET with array fabrication, electronic metrology, the digitization of dose data and automation of the data collection.

Spatial resolution of MOSFET in the “edge-on” orientation appears to match the resolution of Gafchromic film. A further increase of MOSFET resolution is possible if a thinner gate oxide is selected. It should be noted that a reduction in gate oxide thickness reduces the sensitivity of the MOSFET dosimeter. The convenience of remote, on-line measurement and new possibilities of large arrays increases the attractiveness of the MOSFET dosimetry method in measuring of microbeams during MRT and radiosurgery.

Spatial resolution of Gafchromic film is similar to spatial resolution of a MOSFET in the “edge-on” mode for the x-ray beam in this study. Gafchromic film demonstrated a significant lateral nonuniformity of its response to x-ray irradiation. A recently suggested double exposure technique²³ promises to reduce Gafchromic nonuniformity to less than 5%.

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- ³Electronic mail: gkaplan@ucw.edu.au
- ¹A. Holmes-Siedle, "The space-charge dosimeter—General principles of a new method of radiation detection," *Nucl. Instrum. Methods* **121**, 169–179 (1974).
- ²A. B. Rosenfeld, M. G. Carolan, G. I. Kaplan, B. J. Allen, and V. I. Khivrich, "MOSFET dosimeters: role of encapsulation on dosimetric characteristics in mixed gamma-neutron and megavoltage x-ray fields," *IEEE Trans. Nucl. Sci.* **42**, 1870–1877 (1995).
- ³A. Dasu, P.-O. Löfroth, and G. Wickman, "Liquid ionization chamber measurement of dose distribution in small 6 MV photon beam," *Phys. Med. Biol.* **43**, 21–36 (1998).
- ⁴R. C. Hughes, D. Huffman, J. V. Snelling, T. E. Zipperian, A. J. Ricco, and C. A. Kelsey, "Miniature radiation dosimeter for in vivo radiation measurements," *Int. J. Radiat. Oncol., Biol., Phys.* **14**, 963–967 (1988).
- ⁵D. J. Gladstone, X. Q. Lu, J. L. Humm, H. F. Bowman, and L. M. Chin, "A miniature MOSFET radiation dosimeter probe," *Med. Phys.* **21**, 1721–1728 (1994).
- ⁶D. N. Slatkin, P. Spanne, F. A. Dilmanian, and M. Sandborg, "Microbeam radiation therapy," *Med. Phys.* **19**, 1395–1400 (1992).
- ⁷D. N. Slatkin, P. Spanne, F. A. Dilmanian, J.-O. Gebbers, and J. A. Laissue, "Subacute neuropathological effects of microplanar beams of x-ray from a synchrotron wiggler," *Proc. Natl. Acad. Sci. USA* **92**, 8783–8787 (1995).
- ⁸J. L. Laissue, G. Geiser, P. O. Spanne, F. A. Dilmanian, J.-O. Gebbers, M. Geiser, X.-Y. Wu, M. Makar, P. Micca, M. Nawrocky, D. D. Joel, and D. N. Slatkin, "Neuropathology of ablation of rat gliosarcomas and contiguous brain tissues using a microplanar beam of synchrotron-wiggler-generated x rays," *Int. J. Cancer* **78**, 654–660 (1998).
- ⁹W. Zeman, H. J. Curtis, and C. P. Baker, "Histopathologic effects of high-energy-particle microbeams on the visual cortex of the mouse brain," *Radiat. Res.* **15**, 189–495 (1961).
- ¹⁰F. Z. Company and B. J. Allen, "Measurements and Monte Carlo simulations of the fluence and dose characteristics of microplanar proton beam," *Austral. Phys. Eng. Sci. Med.* **19**, 217–223 (1996).
- ¹¹F. Z. Company and B. J. Allen, "Calculation of microplanar beam dose profiles in a tissue/lung/tissue phantom," *Phys. Med. Biol.* **43**, 1–11 (1998).
- ¹²W. R. Nelson, H. Hirayama, and D. W. O. Rogers, "The EGS4 code system," SLAC, Report-265, 1985.
- ¹³A. B. Rosenfeld, G. I. Kaplan, B. J. Zealey, T. Kron, and A. Dilmanian, "Si–SiO₂ interface for characterization of x-ray synchrotron microbeam," in *Materials 98, Proceedings of the Biennial Materials Conference of the Institute of Materials Engineering, Australasia*, edited by M. Ferry (IMEA, Wollongong, 1998), Vol. 1, pp. 401–406.
- ¹⁴A. Holmes-Siedle and L. Adams, "RADFETs: a review of the use of metal–oxide–silicon devices as integrating dosimeters," *Radiat. Phys. Chem.* **28**, 235–244 (1986).
- ¹⁵L. S. August, "Estimating and reducing errors in MOS dosimeters caused by exposure to different radiation," *IEEE Trans. Nucl. Sci.* **NS-29**, 2000–2003 (1982).
- ¹⁶T. Kron, L. Duggan, T. Smith, A. Rosenfeld, M. Butson, G. Kaplan, S. Howlett, and K. Hyodo, "Dose response of various radiation detectors to synchrotron radiation," *Phys. Med. Biol.* **43**, 3235–3259 (1998).
- ¹⁷W. L. McLaughlin, C. Yun-Dong, C. G. Soares, A. Miller, G. V. Dyk, and D. F. Lewis, "Sensitometry of the response of a new radiochromic film dosimeter to gamma radiation and electron beams," *Nucl. Instrum. Methods Phys. Res. A* **302**, 165–176 (1991).
- ¹⁸A. Niroomand-Rad *et al.*, "Radiochromic film dosimetry: Recommendations of AAPM Radiation Therapy Committee Task Group 55," *Med. Phys.* **25**, 2093–2115 (1998).
- ¹⁹L. E. Reinstein and G. R. Gluckman, "Comparison of dose response of radiochromic film measured with He–Ne laser, broadband and filtered light densitometers," *Med. Phys.* **24**, 1531–1533 (1997).
- ²⁰H. E. Johns and J. R. Cunningham, *The Physics of Radiology* (Charles C. Thomas Publishers, Springfield, IL, 1983).
- ²¹D. N. Slatkin, F. A. Dilmanian, M. M. Nawrocky, and P. Spanne, "Design of a multislit, variable width collimator for microplanar beam radiotherapy," *Rev. Sci. Instrum.* **66**, 1459–1460 (1995).
- ²²A. S. Meigooni, M. F. Sanders, and G. S. Ibbott, "Dosimetric characteristics of an improved radiochromic film," *Med. Phys.* **23**, 1883–1888 (1996).
- ²³Y. Zhu, A. S. Kirov, V. Mishra, A. S. Meigooni, and J. F. Williamson, "Quantitative evaluation of radiochromic film response for two-dimensional dosimetry," *Med. Phys.* **24**, 223–231 (1997).