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Dose response of various radiation detectors to synchrotron radiation

Tomas Kron†‡§, Lisa Duggan†, Tony Smith‖, Anatoly Rosenfeld¶, Martin Butson+, Greg Kaplan¶, Steve Howlett† and Kazuyuki Hyodo*

† Newcastle Mater Hospital, Department of Radiation Oncology, Waratah, NSW 2298, Australia
‡ University of Newcastle, Department of Physics, Callaghan, NSW 2308, Australia
‖ University of Newcastle, Division of MRS, Callaghan, NSW 2308, Australia
¶ University of Wollongong, Department of Physics, Wollongong, NSW 2522, Australia
+ Illawarra Cancer Care Centre, Wollongong, NSW 2500, Australia
* Photon Factory, KEK, Tsukuba, Japan

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Abstract. Accurate dosimetry is particularly difficult for low- to medium-energy x-rays as various interaction processes with different dependences on material properties determine the dose distribution in tissue and radiation detectors. Monoenergetic x-rays from synchrotron radiation offer the unique opportunity to study the dose response variation with photon energy of radiation detectors without the compounding effect of the spectral distribution of x-rays from conventional sources. The variation of dose response with photon energies between 10 and 99.6 keV was studied for two TLD materials (LiF:Mg, Ti and LiF:Mg, Cu, P), MOSFET semiconductors, radiographic and radiochromic film. The dose response at synchrotron radiation energies was compared with the one for several superficial/orthovoltage radiation qualities (HVL 1.4 mm Al to 4 mm Cu) and megavoltage photons from a medical linear accelerator. A calibrated parallel plate ionization chamber was taken as the reference dosimeter. The variation of response with x-ray energy was modelled using a two-component model that allows determination of the energy for maximum response as well as its magnitude. MOSFET detectors and the radiographic film were found to overrespond to low-energy x-rays by up to a factor of 7 and 12 respectively, while the radiochromic film underestimated the dose by approximately a factor of 2 at 24 keV. The TLDs showed a slight overresponse with LiF:Mg, Cu, P demonstrating better tissue equivalence than LiF:Mg, Ti (maximum deviation from water less than 25%). The results of the present study demonstrate the usefulness of monoenergetic photons for the study of the energy response of radiation detectors. The variations in energy response observed for the MOSFET detectors and GAF chromic film emphasize the need for a correction for individual dosimeters if accurate dosimetry of low- to medium-energy x-rays is attempted.

1. Introduction

Radiation dosimetry is one of the most important tasks in radiation therapy and health physics. As the aim of dosimetry in these fields is to predict biological outcomes, the dose should be known for water or human tissues. However, most radiation detectors are not tissue equivalent. While this is not of major concern for megavoltage photons, problems can arise at low- to medium-energy photons. This is due to the strong dependence of the photoelectric effect on the effective atomic number of the material in which energy

§ Address for correspondence: Department of Radiation Oncology, Newcastle Mater Hospital, Locked Bag 7, Hunter Region Mail Centre, NSW 2310, Australia. E-mail address: phtk@cc.newcastle.edu.au

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is deposited by the ionizing radiation (Johns and Cunningham 1983). In addition the dose response of some radiation detectors such as thermoluminescence dosimeters (TLDs) depends on the spatial distribution of the energy deposition events or the lineal energy (Horowitz 1984). For x-rays the variations are largest in the kilovoltage energy range. The resultant variation of dose response with photon energy is difficult to assess as conventional radiation sources such as the anodes of x-ray tubes emit a spectrum of different x-ray energies. Monoenergetic photons extracted from synchrotron radiation offer the unique opportunity to study the energy response of radiation detectors without the compounding effect of the spectral distribution of x-rays from conventional sources.

It was the aim of the present study to evaluate the variation of dose response with photon energy for a variety of radiation detectors. The detectors studied were radiographic film, radiochromic film, thermoluminescence dosimeters (TLDs) and metal oxide semiconductor field effect transistors (MOSFETs). The knowledge of the performance of these detectors with synchrotron radiation is also a topical matter as synchrotron radiation is increasingly used in a variety of medical applications (Burattini *et al* 1995, Chapman *et al* 1997, Lewis 1997). The dosimeters included in the present study are listed in table 1. A plane parallel ionization chamber specifically designed for low-energy photons (NE2532/2) was used as reference.

Table 1. Radiation detectors used in the study.

Radiation detector	Type	Comments
Ionization chamber	NE2532/2	Perspex plane parallel chamber, 0.02 cm^3 volume, used as reference dosimeter
Radiographic film	Kodak X-Omat V	Radiotherapy verification film, prepacked, developed at the Newcastle Mater Hospital
	CEA verification film	Scandinavian film with linear dose response due to homogeneous grain size (Cheng and Das 1996). This film was not studied in synchrotron radiation
Radiochromic film	GAF chromic film batch: A: MD-55 B: MD-870, NA 37-040 C: NA 37-041	GAF chemicals, evaluated at Illawarra Cancer Care Centre using a modified Scantitronix film scanner
Thermoluminescence dosimeters (TLDs)	LiF:Mg, Ti	Standard TLD material Harshaw TLD 100, a fast cool down annealing process was used
	LiF:Mg, Cu, P	High-sensitivity TLD material obtained from the Solid Dosimeter Lab in Beijing
MOSFET detectors	Wollongong University	Manufactured for the present study by the University of Wollongong. Detectors embedded in solid water phantom material (compare figure 2). Sample A and B: Covered with 20 μm protective polystyrene cover; sample C: No protective cover (compare Butson *et al* 1996b)
	TN-RD-50, two detectors	Commercially available TLD system from Thomson and Nielsen for *in vivo* dosimetry. These detectors were not studied in synchrotron radiation beams

1.1. Radiographic film

Due to its excellent spatial resolution in two dimensions, film dosimetry is a useful tool for radiation detection and dosimetry in radiotherapy (Williamson *et al* 1981, Mota *et al*

1990). However, radiographic film is based on the reduction of silver halide (usually silver bromide on a gelatine base) due to ionizing radiation. The resulting effective atomic number depends on the silver bromide loading and is considerably higher than that of tissue or water. This leads to an overresponse of film to low-energy x-rays as demonstrated by Muench *et al* (1991). The energy dependence of film is complicated by the fact that the range of secondary electrons from low-energy x-rays is of the order of 10 μm in tissue (Siemens 1993). This is comparable to the typical grain size in film (Dutreix and Dutreix 1969, van Battum and Huizenga 1990, Cheng and Das 1996) which could result in a dependence of the dose response on grain size and distribution.

More recently a new type of radiographic film, CEA film, has been introduced (Cheng and Das 1996). This film features uniform silver halide grain size which results in a linear dose response curve. Samples of this film became available to us after the synchrotron experiments. Therefore, it was only studied in spectral x-rays from a superficial/orthovoltage treatment unit.

1.2. Radiochromic film

Radiochromic film consists of a polyester base with one or two thin radiochromic layers which turn blue on irradiation without the need for developing (Saylor *et al* 1988, McLaughlin *et al* 1991, Muench *et al* 1991, Butson *et al* 1996a). Muench *et al* (1991) studied the variation of dose response of GAF chromic film with equivalent photon energy of a superficial/orthovoltage unit and found a significant underresponse at an equivalent photon energy of 28 keV. The present study is limited to single-emulsion film and does not evaluate the more recent film MD-55-2 with increased sensitivity, which features two layers of radiochromic emulsion.

1.3. Thermoluminescence dosimeters

Thermoluminescence dosimeters (TLD) are often categorized in high- and low-Z materials (McKinlay 1981, Kron 1994, McKeever *et al* 1995). Lithium fluoride is the most commonly used TLD material in medicine and one of the most often quoted advantages of this low-Z TLD is its tissue equivalence (Horowitz 1984, Kron 1994). In the present study two types of this TLD material were investigated: lithium fluoride doped with magnesium and titanium (LiF:Mg, Ti) and a more recent development from the same family of materials, LiF:Mg, Cu, P (Wu *et al* 1984, Zha *et al* 1993, McKeever *et al* 1991, Horowitz 1993, Delgado *et al* 1995). The latter is approximately 30 to 50 times more sensitive than LiF:Mg, Ti and exhibits less supralinear behaviour.

The dose response of LiF:Mg, Ti to synchrotron radiation has been previously studied by Nariyama *et al* (1993) and the same team has recently also investigated lithium borate, beryllium oxide and calcium sulphate in synchrotron radiation (Nariyama *et al* 1997). Another interesting approach to the study of dose response as a function of photon energy has recently been described by Edwards *et al* (1997). Amongst other solid state radiation detectors, they studied the variation of dose response in LiF:Mg, Ti rods in quasimonoenergetic x-rays produced by a Pantak HF-320 x-ray unit. These quasimonoenergetic beams were produced by appropriate choice of filtration and typically feature a mean energy of 80% of the peak energy with a full width half maximum (FWHM) of some 30% of the mean energy.

Both lithium fluoride materials together with aluminium oxide TLDs have been subject to a previous study by our group (Kron *et al* 1996) where LiF:Mg, Cu, P was found to be

more tissue equivalent than LiF:Mg, Ti. The present study was to build on these preliminary experiences which will be discussed below.

1.4. Metal oxide semiconductor field effect transistors (MOSFETs)

The operation of MOSFET dosimeters is based on the build up of charge created by ionizing radiation in the gate silicon oxide. This build up of charge is responsible for a shift of the threshold voltage of the MOS field effect transistor to a lower voltage, which is a measure of absorbed dose in the gate oxide. As extremely small detectors (thickness approximately 1 μm), MOSFET detectors have recently found a number of applications in medicine ranging from *in vivo* dosimetry to surface dose measurements (Soubra *et al* 1994, Butson *et al* 1996b, Rosenfeld *et al* 1997).

The mechanism of variation of energy response of the MOS structures has been studied in the energy range 70 eV–20 keV with particular interest in the range 1–20 keV (Dosier and Brown 1981, Humm 1986). In this photon energy range the energy response was found to be determined by the dynamics of recombination of the electron–hole pairs in silicon oxide. The increasing recombination effects with decreasing photon energy result in a reduced sensitivity. Other contributing factors are the predominance of the photoelectric effect in SiO_2 in comparison with tissue for photon energies less than 100 keV, a dose enhancement effect due to the metal–SiO_2 interfaces used in the MOSFET devices (Fleetwood *et al* 1996) and the packaging of the chip. This packaging effect was investigated using spectral x-ray units for photons between 20 kV and 6 MV (Bruker *et al* 1995, Rosenfeld *et al* 1995) and could account for a dose enhancement of up to a factor of 18 if materials with high effective atomic numbers are used in the packaging. In the present study experimental MOSFET detectors were used similar to the ones described for surface dose measurements in megavoltage photon beams by Butson *et al* (1996b).

MOSFET detectors are now also commercially available for radiotherapy dosimetry (Thomson and Nielson, Ottawa, Canada). These detectors are dual-bias dual-MOSFETs and designed for clinical dosimetry. T&N MOSFETs have been studied by Soubra *et al* (1994) and their energy response was recently evaluated by Edwards *et al* (1997) in quasimonoenergetic x-ray beams. A set of these detectors was given to our institution for evaluation. Unfortunately, they only became available after the synchrotron radiation beam time and were as such only studied in spectral x-rays.

2. Materials and methods

2.1. Sources of x-rays

Table 2 lists all radiation qualities included in the present study. They included both monoenergetic photons filtered from synchrotron radiation and spectral x-rays from conventional sources such as x-ray tubes and medical linear accelerators. The synchrotron radiation beams from beamline NE5A were only applied to TLDs and the results of this study have been reported previously (Kron *et al* 1996). As the present study was an extension of the experiments performed at NE5A, the beams are included here for completeness.

2.1.1. Synchrotron radiation. The synchrotron radiation experiments were performed over two separate weeks at the BL14C experimental station in the photon factory and the station NE5A at the accumulation ring, both at the National Laboratory for High Energy Physics (KEK), Tsukuba, Ibaraki, Japan.

Table 2. Summary of the radiation beams used in the present study—the synchrotron radiation beams used at beamline NE5A marked with # have been used in a previous study on TLDs only (Kron *et al* 1996) (S = synchrotron, BL = beam line, DCM = dual-crystal monochromator, ADM = asymmetric diffraction monochromator, linac = medical linear accelerator).

Effective energy (keV)	Peak energy (keV)	Produced by	HVL (mm)	Field size (cm^2)	Approximate dose rate (Gy min^{-1})	Comments
10	10	S, BL 14C, DCM	0.09 Al	2.5×1.4	18	
15	15	S, BL 14C, DCM	0.31 Al	2.5×1.0	15	
20	20	S, BL 14C, DCM	0.62 Al	2.5×0.9	3	
26	26	S, BL 14C, DCM	1.5 Al	2.5×0.5	2	Reference beam
26.8#	26.8	S, BL NE5A, ADM	1.8 Al	9×2	2	
33.2#	33.2	S, BL NE5A, ADM	2.9 Al	7.5×2.5	2	
40#	40	S, BL NE5A, ADM	4.6 Al	7.5×6	3	
80.4#	80.4	S, BL NE5A, ADM	1.6 Cu	9×1.5	0.4	Inhomogeneous beam, third harmonics
99.6#	99.6	S, BL NE5A, ADM	1.9 Cu	7.5×1.5	0.4	Third harmonics
35	60	Siemens Stabilipan II superficial beam	1.6 Al	Cone 4 cm diameter	1	Spectrum
47	120	Siemens Stabilipan II superficial beam	2.8 Al	Cone 4 cm diameter	3	Spectrum
72	180	Siemens Stabilipan II orthovoltage beam	0.5 Cu	10×8 diameter	1.1	Spectrum
112	250	Siemens Stabilipan II orthovoltage beam	2 Cu	10×8	0.9	Spectrum
165	300	Siemens Stabilipan II orthovoltage beam	4 Cu	10×8	0.5	Spectrum
1300	4000	Varian Clinac 600C, linac	11 Pb	10×10	2.5	Spectrum
2000	6000	Varian Clinac 1800, linac	12 Pb	10×10	4	Spectrum
6000	18 000	Varian Clinac 1800, linac	12 Pb	10×10	4	Spectrum

At beamline BL14C monoenergetic photons were produced from 'white' synchrotron radiation using dual diffraction from the (1,1,1) plane of a silicon crystal. The energy was tuned by altering the incidence and reflection angle of the x-rays onto the crystal. The spectrum of the white x-rays at the experimental station has a broad maximum of x-ray intensity at 15 keV falling to less than 10% of this at 60 keV. The energies selected are given in table 2. The full width half maximum of the energy distribution was less than 1% of the selected energy. The experimental set-up is illustrated in figure 1. The radiation detectors were placed on a movable carriage, which allows the detector to be moved through the radiation field. The carriage was moved by a step motor with a minimum step size of 10 μm controlled by a computerized controller (AS NET-3, Sigma Koki Co., Japan).

The radiation beams at beamline NE5A used in the previous study were produced using asymmetric diffraction from a silicon (3,1,1) plane (Hyodo *et al* 1991, Ando *et al* 1993). This allows a larger radiation field to be obtained. The disadvantage of this procedure is a relatively inhomogeneous radiation field and the fact that the beam is contaminated with the third harmonics of the beam energy. This was utilized to produce monoenergetic photon beams of higher energy by filtering the main radiation component and utilizing only the

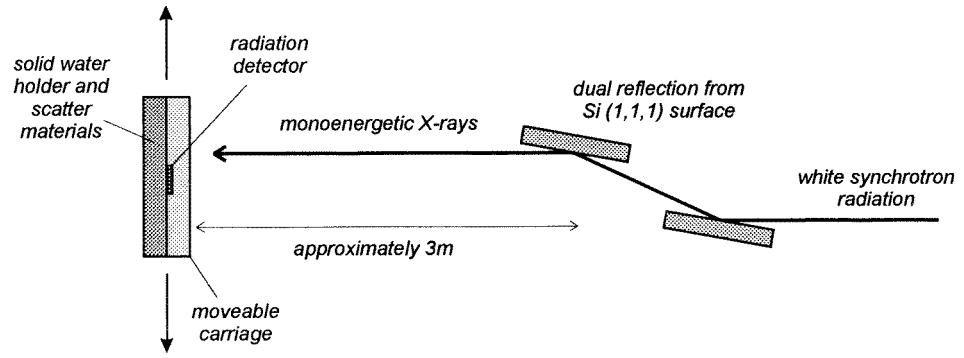

Figure 1. Schematic drawing of the experimental set-up for irradiation of detectors using synchrotron radiation.

third harmonics. The 80.4 keV and 99.6 keV beams listed in table 2 were obtained in this fashion (Kron *et al* 1996). In these experiments the detectors could only be placed stationary in the beam.

The half-value layer (HVL) for each beam was measured using high-purity aluminium plates and the plane parallel ionization chamber (PTW 2532/2). The measurements were performed with the ionization chamber in air. At least the first three HVL were measured in order to assess the contamination of the photon beam with higher energy radiation. The results of these measurements are given in table 2.

The field size of the different radiation beams was not equal due to diffraction angle and crystal size. It was collimated to a rectangular useful beam and assessed using both film types and by moving the plane parallel ionization chamber through the beam. The output was determined at the centre of the field using the plane parallel ionization chamber. The radiation dose was controlled by the time the shutter was open. A timer error allowing for the shutter speed was corrected for.

The calculated dose rate and field size of all synchrotron radiation beams is listed in table 2. The dose rate was not constant as it varies with the current in the storage ring. This was corrected for by using the ionization chamber some 2 cm behind the investigated detectors as dose monitor in the same field (beamline BL14C). In the accumulation ring (beamline NE5A) the dose was calculated from measurements before and after the TLD irradiation. In both experimental set-ups the current in the storage ring was recorded and used to verify the variation in output.

2.1.2. Conventional spectral x-rays. The dose response at synchrotron radiation energies was compared with the one for several superficial/orthovoltage radiation qualities (HVL 1.4 mm Al to 4 mm Cu, Siemens Stabilipan II) and 4, 6 and 18 MV photons from a medical linear accelerator (Varian Clinac 600C/4MV and Varian Clinac 1800). The characteristics of the superficial/orthovoltage unit are similar to the ones described by Niroomand-Rad *et al* (1987).

The effective energy of the low- to medium-energy x-rays was determined from the accelerating voltage and the first and second HVL of the radiation beams using a method described by Seelentag and Panzer (1980). It is given in the first column of table 2. The detectors were exposed at the surface of a solid water phantom at normal treatment distance with the exception of the radiographic film which was exposed at extended focus to surface

distance to improve the accuracy of the dose determination for small radiation doses. A thin layer of plastic foil ('glad wrap') was used on the lead glass cones in the 60 and 120 kVp beams to absorb contamination electrons (Klevenhagen *et al* 1991, Ferguson *et al* 1995).

The spectral composition of the megavoltage photon beams is difficult to ascertain. Its effective energy is approximately 1.5 MeV and 2 MeV for the 4 MV and 6 MV beam respectively. Typical spectra are given, for example, by Mohan and Chui (1985). The detectors were exposed at the depth of maximum dose in a solid water (RMI) slab phantom. Throughout the study, this material (Constantinou *et al* 1982) was used as the phantom material.

2.2. Radiation detectors

2.2.1. Ionization chamber. A Perspex parallel plate ionization chamber (NE2532/2) was used as the dose reference in the synchrotron radiation beams of beamline BL14C. This chamber is specifically designed for superficial radiation beams and features a thin entrance window (2.5 μg cm^{-2}) of graphite-coated polyethylene, an active volume of 0.02 cm^3 and a plate separation of 1 mm (Mitchell *et al* 1998). Calibration factors were obtained for the chamber by intercomparison in the superficial beams (60 kVp and 120 kVp) of the Siemens Stabilipan II unit. The calibration of this unit is done using the protocol of the Institute of Physical Sciences in Medicine (IPSM 1991). This was done using a calibrated Farmer-type cylindrical ionization chamber (NE2571) and the calibration is traceable to the Australian Radiation Laboratory (ARL) in Melbourne. The NE2532/2 chamber was used with a Victoreen 500 electrometer.

The radiation beams of the NE5A synchrotron radiation station were calibrated using a PTW 23344 thin window parallel plate ionization chamber of volume 0.2 cm^3. More details of this procedure are given in Kron *et al* (1996). The superficial/orthovoltage unit was calibrated using the IPSM (1991) protocol and the medical linear accelerators according to the protocol of the International Atomic Energy Agency (IAEA 1987).

2.2.2. Radiographic film. The radiographic film (Kodak X-Omat V) was taken from Newcastle, Australia to the synchrotron radiation experimental stations. In all experiments it was exposed in its envelope without moving the film in the radiation beam. The beam was left on for the required amount of time and the dose delivered at BL14C was monitored using the parallel plate ionization chamber mounted 5 mm behind the film. To conserve film more than one exposure was made on the same sheet of film during the synchrotron radiation experiments at BL14C where the radiation fields were small (compare table 2).

All films were taken from the same batch of film. They were developed together using an automatic processor (Agfa Curix). The optical density, OD, of the films was evaluated using a film densitometer (Wellhoefer). Base and fog was corrected for using an unexposed film of the same batch.

Towards the end of the study a novel type of film became available. CEA treatment verification film (TVS/EP, CEA, Sweden) was exposed in its characteristic shiny vacuum sealed polyester envelope. The envelope is approximately 130 μm thick (Cheng and Das 1996). It was exposed to superficial/orthovoltage radiation from the Siemens Stabilipan II unit and megavoltage x-rays from the medical linear accelerators only. The linear dose response relationship, which is described in the literature (Cheng and Das 1996) was verified in the 60 kVp beam only.

2.2.3. Radiochromic film. Three sheets of radiochromic film (GAF-chemicals, models MD-55, lot no 890-5 = film A, MD-870, NA model 37-040 = film B and 37-041 = film C) were investigated. The MD-55 film was single-emulsion only as opposed to the more recently developed double-emulsion film MD-55-2 (Meigooni *et al* 1996). All films were cut into small pieces of approximately 15×50 mm^2 and irradiated individually. The relative optical density was determined using a Scanditronix film densitometer at the Illawarra Cancer Care Centre. The densitometer was modified by adding an ultrabright 3000 mcd light emitting diode (LED) with a peak wavelength of 660 nm and a spectral width of 10 nm (FWHM).

2.2.4. Thermoluminescence dosimeters. LiF:Mg, Ti TLD ribbons (Harshaw TLD 100, Solon Ohio, USA) of $3.1 \times 3.1 \times 1$ mm^3 and LiF:Mg, Cu, P (GR-200, Solid State Dosimeter Laboratory, Beijing, China) of 4.5 mm diameter and a thickness of 0.8 mm were included in the study. All TL detectors were stored and annealed in aluminium trays and handled with vacuum tweezers. Annealing was performed in a muffle oven (Isuzu AT-S13) with a temperature overshoot of up to 20 °C over 15 min. This was reproducible within 5 °C and the annealing temperature was set 10 °C lower than the desired temperature to account for this problem (for example 390 °C to achieve a nominal temperature of 400 °C). The LiF:Mg, Ti ribbons were annealed for 30 min on 400 °C. Then the tray was taken out of the oven and placed on an aluminium block kept at room temperature. Using this procedure, the TL detectors were cooled down from 400 °C to less than 100 °C in less than 3 min.

The LiF:Mg, Cu, P were annealed for 15 min at 270 °C and then fast cooled down as described above. However, the relatively short time the detectors were exposed to a temperature above 270 °C had already affected the sensitivity of the material dramatically as previously described by Chandra *et al* (1982) and Furetta *et al* (1994). A sensitivity loss of approximately a factor of 50 was observed which also worsened the reproducibility of the results. This is reflected in the larger error bars in figure 8. The results of the first experimental run at beamline NE5A were not affected by this as the TLDs were prepared in Newcastle with a more suitable annealing oven (STM) (Kron *et al* 1996).

After irradiation the TL detectors were read out in a manual TLD reader (Victoreen 2800M) using a two-step readout procedure (LiF:Mg, Ti step 1: 10 s 160 °C; step 2: 20 s 300 °C; LiF:Mg, Cu, P step 1: 10 s 135 °C; step 2: 30 s 240 °C). The glow curve was checked on the screen of the reader, but no computerized glow curve analysis was performed due to experimental limitations. The TL emission was evaluated using regions of interest (ROIs) in the glow curve which were defined covering glow peaks of interest. In most cases, the delay between irradiation and readout was kept below 12 h.

The physical TLD thickness of nearly 1 mm results in x-ray attenuation already throughout the detector. This self-absorption is particularly noticeable in the lower-energy x-rays and was corrected for following a method described by Nariyama *et al* (1993, 1997). If one assumes exponential dose fall off within the detector of thickness d, one can calculate the correction factor F from the photon attenuation coefficients μ_{att} as

$$F = [1 - \exp(-\mu_{\mathrm{att}} d)]/(\mu_{\mathrm{att}} d). \tag{1}$$

F is a function of the photon energy and values for μ_{att} can be found for example in Johns and Cunningham (1983). The formula assumes no changes of energy spectrum throughout the detector and ignores effects at the edge of the detector.

In addition to irradiation with synchrotron radiation the TL dosimeters were irradiated three times in a 10×10 cm^2 radiation field of a spectral 6 MV photon beam from a medical linear accelerator. In these irradiations the TL dosimeters were exposed to 200 mGy in

a solid water (RMI) phantom at the depth of maximum dose (15 mm) at 100 cm focus to phantom surface distance. The absolute dose delivered was accurate within ±2% of the stated dose. The three irradiations with 6 MV spectral x-rays were performed after the synchrotron radiation experiments and were used for two purposes: (i) to obtain an estimate of the reproducibility of the reading for each TL detector type, and (ii) to establish a radiation sensitivity factor for each individual TL dosimeter. These factors were used as calibration factors for individual dosimeters to evaluate all other results (Kron *et al* 1993).

2.2.5. MOSFET detectors. The MOSFET dosimeter system used in the present study was developed by the radiation physics group at the University of Wollongong (Australia). The n-MOSFET detectors, which were obtained from the Ukraine, feature a 1 μm thick silicon oxide layer with an aluminium gate electrode of 0.5 μm thickness (Rosenfeld *et al* 1995). For these experiments the detector was placed in a solid water plate of approximately 20×40 mm^2 and a thickness of 4 mm. The detector assembly is shown in figure 2. The surface of the chip was lined up with the solid water surface. As can be seen in table 1 two of the three chips used in the present study were covered with a thin layer of polystyrene of about 60 μm thickness.

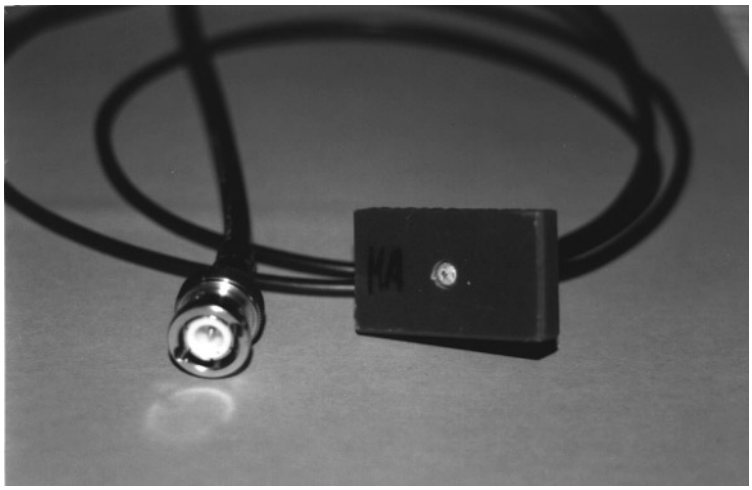

Figure 2. Photograph of the Wollongong-type MOSFET detector A (compare table 1) embedded in a solid water holder. The active area is in the centre of the circular area. It is covered with a 60 μm thick polystyrene cover.

The gate of the MOSFETs was biased during irradiation with 5 V. They were evaluated using a battery operated portable reader which was designed for measurements of the threshold voltage in the range of +10 V to −10 V. The stability of the threshold voltage assessment was within 2 mV.

In addition to this a commercial system (TN-RD-50, Thomson and Nielson, Ottawa) was tested. These detectors only became available after the synchrotron radiation experiments and were only studied in spectral x-rays. The detectors are mounted at the end of a flexible band covered with 2 mm epoxy on one side. They were exposed with the epoxy build-up facing the radiation beam. Two identical detectors were studied.

All radiation detectors used in the present study are listed in table 1.

2.3. Evaluation

2.3.1. Dose response. The dose response R was fitted using a single exponential fit

$$R = R_{\max}[1 - A\exp(-aD)] \tag{2}$$

with $R_{\max}$ as the maximum response and D the dose delivered. The parameters A and a are fitting parameters. This fit was used with three parameters ($R_{\max}$, A and a) for all types of film and the LiF:Mg, Cu, P thermoluminescence dosimeters.

The film response can be expressed by the optical density OD. Two modifications of the fit were tried for radiographic film. A two-parameter fit with $OD_{\max}$ fixed to 4 did not fit the data sets well where no exposures were available with optical densities larger than 2 (compare figure 3). Similarly, a two-exponential fit did not improve the results significantly and tends to overparametrize the experimental data available.

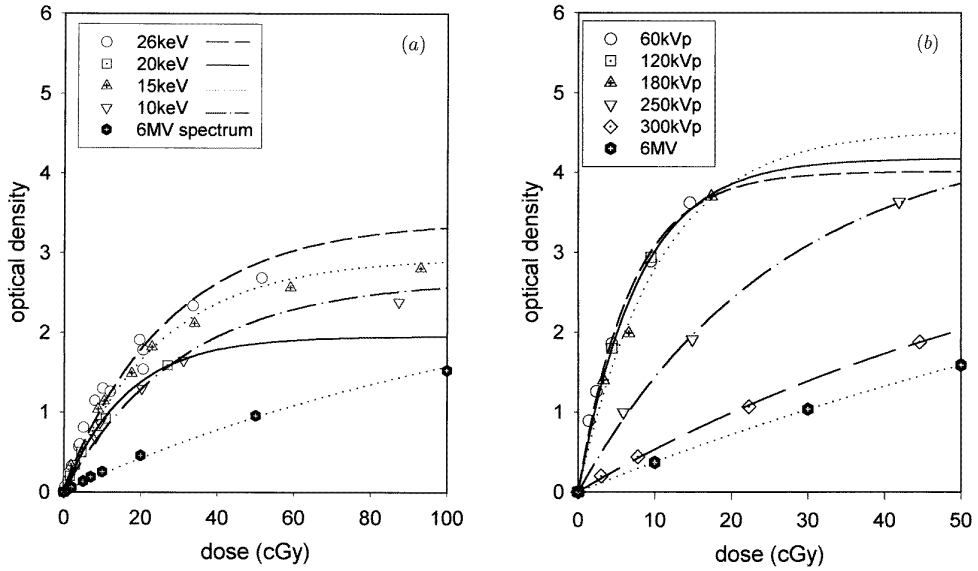

Figure 3. Dose response of radiographic film exposed to different radiation qualities. The curves represent the best fit using the monoexponential function given in equation (2). (*a*) Exposure to synchrotron radiation. The difference in saturation density is an artefact arising from the fact that the doses given were insufficient to cause optical density saturation in all cases (particularly in the case of 20 keV). (*b*) Exposure to spectral x-rays from a superficial/orthovoltage unit and a medical linear accelerator. The dose response to 6 MV spectral x-rays from the linear accelerator is shown in both graphs for comparison.

The dose response in a particular experiment was then determined from the slope of the dose response curve at a particular physical response. The choice of this response was not critical for the TL detectors and the CEA film studied as the dose response for both detector types was found to be virtually linear over the dose range of interest. For the other detector types the dose response was specified at the following physical response levels:

(i) radiographic film—optical density of 1;
(ii) radiochromic film—optical density of 0.3;
(iii) MOSFET detectors—threshold voltage 0.

The choice of an equal dose level for the specification of dose response (rather than an equal response level) was checked. The results obtained this way are qualitatively identical to the results presented below.

2.3.2. Variation of response with energy. A simple model was devised to describe the variation of detector response R with energy E. It is based on the assumption that the response at low energies is reduced exponentially while it varies at medium energies with the energy dependence of the photoelectric effect. For materials with low atomic number, the cross section for photoelectric effect is approximately inversely proportional to the cube of the photon energy (Johns and Cunningham 1983, Khan 1994, Metcalfe *et al* 1997). Therefore, one can write

$$R(E) = \{1 - \exp[-\alpha_1(E - E_1)]\}[1 + \alpha_2/(E - E_2)^3] \tag{3}$$

with α_1 and α_2 as two fitting parameters which determine the importance of exponential fall-off towards low energies and inverse cubic fall-off towards higher energies respectively. The other two fitting parameters E_1 and E_2 allow for an energy shift for the two components. The maximum response depends on all four parameters.

The model is appropriate for a dose response ratio (for example relative to water) which is normalized to a high photon energy, such as ^{60}Co or megavoltage photons from linear accelerators. In the present study all dose responses were normalized to the response to 6 MV spectral x-rays from a Varian Clinac 1800 medical linear accelerator. The fit does not require the knowledge of any interaction coefficients and does not allow one to estimate them.

The fitting was performed on an IBM compatible personal computer using SigmaPlot 3.0 (Jandel Scientific) software. It was used for radiographic film and the two solid state dosimeter types (TLD and MOSFET). No satisfactory fit was reached in the case of the radiochromic film.

3. Results

3.1. Radiographic film

Figure 3 shows the dose response of Kodak X-Omat V film to various x-ray qualities. Figure 3(*a*) depicts the response to synchrotron radiation from BL14C while figure 3(*b*) shows the dose response following irradiation with superficial/orthovoltage beams. For comparison, the dose response for a 6 MV megavoltage photon beam is also shown in both graphs. The lines represent the best fit according to equation (2).

Figure 4 depicts the variation of dose response of film with photon energy. This was calculated from the dose required to produce an optical density of 1. The lines show the best fit according to equation (3). The maximum response of Kodak X-Omat V was found to be 12 times the response of water at an effective energy of 39 keV. This fit includes all data points. In order to allow a direct comparison with the CEA film where no synchrotron results were available, the fit was repeated with spectral radiation beams only (not shown in figure 4). A maximum response of 15.7 and 25.3 was found at 41 keV and 42 keV for X-Omat V and CEA film respectively. The dose response of radiochromic film is also shown in figure 4 for comparison.

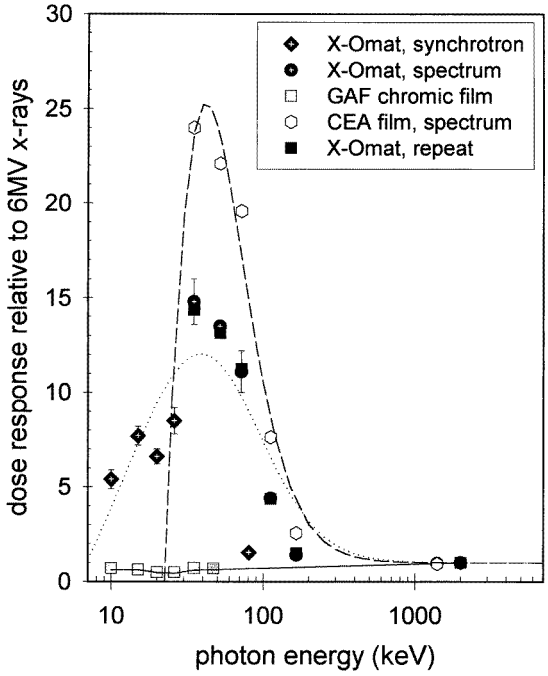

Figure 4. Variation of the dose response with effective radiation energy for radiographic film. The response is normalized to 1 for 6 MV spectral x-rays from a medical linear accelerator. The curves represent the best fit according to equation (3) to all data (synchrotron and spectral beams in the case of X-Omat V). For comparison the variation of dose response for radiochromic film (MD-55) is also given. This is shown in more detail in figure 6.

3.2. Radiochromic film

Radiochromic film is approximately 100 times less sensitive than standard radiographic verification film. The dose response for radiochromic film as measured using the converted Scanditronix densitometer proved to be nonlinear, as can be seen in figure 5. This figure depicts the dose response of GAF chromic film MD-55 (A) and MD-870 (B) to monoenergetic photons of 26 keV. The simple monoexponential model of equation (2) proved to be a reasonable fit for exposures not reaching saturation. The sensitivity of the MD-55 film proved to be approximately 3.2 times higher than that of MD-870 film. Due to the limited number of samples of GAF chromic film, no dose response curve was acquired for the other radiation qualities. Here, the film was always exposed to a dose which was estimated to yield an optical density of approximately 0.3.

The variation of dose response was determined from the dose required to achieve an optical density of 0.3 in radiochromic film. Figure 6 shows the variation of sensitivity with energy relative to the dose response to 6 MV x-rays. The radiochromic film underestimated the dose at 25 keV photon energy by approximately a factor of 2. As can be seen in figure 6, the three types of radiochromic film exhibited a similar underresponse at these energies even if their actual sensitivity varies by more than a factor of 3.

The simple model of equation (3) did not adequately describe the observed underresponse at low energies relative to water. It would require an additional term in the equation to describe the observed behaviour. As the four-parameter fit of equation (3)

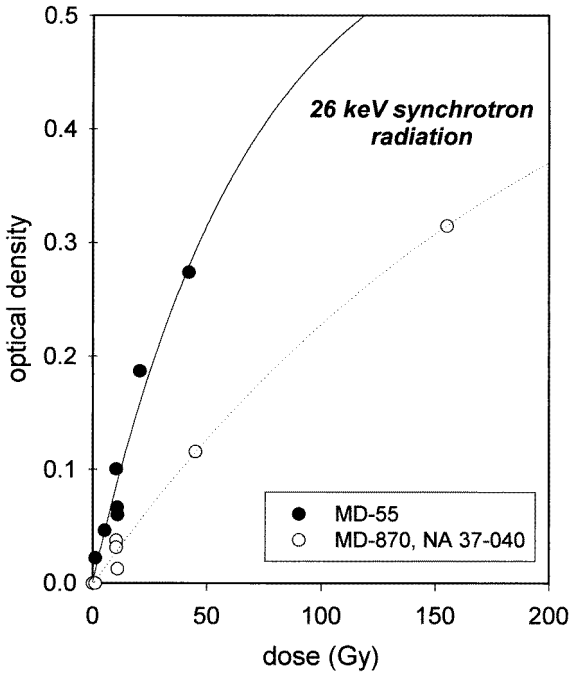

Figure 5. Dose response of radiochromic film MD-55 and MD-870 to 26 keV x-rays from beamline BL 14C. The lines represent the best fit using the monoexponential function of equation (2).

overparametrizes the experimental data already, the minimum responses were estimated from the experimental data. They are listed with those from the other detectors in table 3.

For comparison the ratio of mass energy absorption coefficients for GAF chromic film relative to water is shown in figure 6 as well. The data that were taken from McLaughlin *et al* (1991) describe the experimental findings qualitatively even if the magnitude of the effect appears to be larger than predicted by the energy absorption coefficients alone.

3.3. Thermoluminescence dosimeters

Figure 7 shows the dose response for LiF:Mg, Cu, P TLDs exposed to monoenergetic photons of 26 keV. The response is linear up to some 30 Gy. These results are not affected by the loss of sensitivity due to the inappropriate annealing described in section 2.2.4 as this set of TLDs was annealed and read out in Newcastle. The uncertainty in the measurements is mainly due to uncertainty in the determination of the delivered dose due to the very long times required to deliver radiation doses of several hundred grays (>10 h). In this time the electron storage ring current varied significantly and leakage in the ionization chamber/electrometer reference dosimeter became noticeable.

The monoexponential model of equation (2) describes the data reasonably well. A dual-exponential approach would have improved the fit, but more experimental data around 150 Gy would have been required to justify this. Saturation is reached at approximately 150 Gy. Within the uncertainty of the experiment no supralinearity could be observed. The acquisition of a dose response curve was not repeated for other radiation qualities due to the long irradiation times required to reach saturation dose.

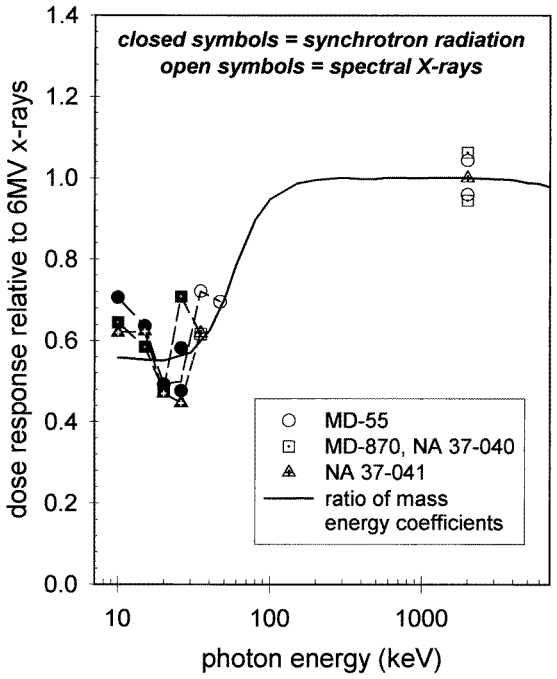

Figure 6. Variation of the dose response with effective radiation energy for radiochromic film. The response is normalized to 1 for 6 MV spectral x-rays from a medical linear accelerator. The full curve was calculated from the ratio of mass energy absorption coefficients between GAF chromic film and water as given by McLaughlin *et al* (1991).

Figure 8 shows the variation of dose response with radiation quality for the two TLD materials investigated. The synchrotron results were corrected for self-absorption according to equation (1). The large uncertainties of the synchrotron measurements for LiF:Mg, Cu, P are a result of the loss of sensitivity as a result of inappropriate annealing as described in section 2. The lines represent the best fit of the experimental data using equation (3). The data point obtained for 80.4 keV at beam line NE5A was not included in the fit. As discussed in Kron *et al* (1996), the beam inhomogeneity appears to be have been too large to get reliable dosimetric results.

The overresponse of LiF:Mg, Ti was found to be 1.47 at 27 keV photon energy. It is significantly higher for low-energy photons than the one for LiF:Mg, Cu, P which was found to be only 5% higher than unity at 24 keV. The fit for LiF:Mg, Cu, P also models the anomalous energy response of the material at 80 keV (Olko *et al* 1993). The present fit estimates a dose response minimum of 0.8 at approximately 200 keV. The fact that the energy of the minimum is too high is a result of the insufficient experimental data above 100 keV.

3.4. MOSFETs

Figure 9 shows the reduction of dose response in the Wollongong MOSFET A as a function of total gate voltage. The total gate voltage is related to the total absorbed dose that is also

Table 3. Summary of the extrema in dose response found for different radiation detectors. Except for the radiochromic film all data were derived using the fit of equation (3).

Radiation detector	Max/min	Effective energy (keV)	Relative response	Comment
Radiographic film				
X-Omat V	Maximum	39	12.0	Fit all energies
	Maximum	41	15.7	Fit only spectral beams
CEA film	Maximum	42	25.3	Fit only spectral beams
Radiochromic film				
A: MD-55	Minimum	24	0.48	Estimated from graph
B: MD-870, NA 37-040	Minimum	21	0.48	Estimated from graph
C: NA 37-041	Minimum	26	0.45	Estimated from graph
Thermoluminescence dosimeters (TLDs):				
LiF:Mg, Ti	Maximum	27	1.47	Fit from all energies
LiF:Mg, Cu, P	Maximum	24	1.05	Fit from all energies
MOSFET detectors				
Wollongong 'B'	Maximum	34	7.1	Fit from all energies
Wollongong 'C'	Maximum	36	6.5	Fit from all energies
Thomson and Nielsen (T&N) 1	Maximum	56	3.1	Fit only spectral beams
T&N 2	Maximum	51	3.3	Fit only spectral beams
Data from Edwards *et al* (1997)				Fit data from literature
TLD rods LiF:Mg, Ti	Maximum	32	1.32	Covered with 0.3 mm PVC
TLD rods LiF:Mg, Ti with thin film	Maximum	27	1.28	Covered with 0.1 mm Nescofilm
Diode EDP 10	Maximum	105	1.62	0.75 mm steel build-up
Diode EDD 5	Maximum	40	6.43	4.5 mm Perspex build-up
MOSFET T&N '1'	Maximum	32	4.0	Under 2 mm epoxy
MOSFET T&N '2'	Maximum	31	4.3	Under 2 mm epoxy

indicated on the horizontal axis. The reduction of dose response, expressed as gate voltage change per absorbed dose, is a linear function of the total absorbed dose. The detector response is reduced by approximately 3% for a gate voltage change of 1 V. This function was used to correct the readings of the other Wollongong MOSFET detectors.

Figure 10 shows the variation of dose response with photon energy relative to water for MOSFET detectors. The curves show the best fit using equation (3). The two Wollongong-type detectors exhibit a maximum of dose response relative to water around 35 keV. The difference between the two detectors in the absolute dose maximum of 7.1 and 6.5 respectively approaches statistical significance.

For comparison the results obtained in spectral radiation beams with the Thomson and Nielson MOSFET detectors are also plotted in figure 10. They have only been studied in the spectral beams from the superficial/orthovoltage unit. The reduction of dose response with total absorbed dose was found to be less than 1% for a gate voltage change of 1 V which is considerably smaller than the one observed in the Wollongong MOSFETs. The dose response for different radiation energies was corrected for this variation.

As can be seen in figure 10, the Thomson and Nielson detectors also exhibit a smaller variation of dose response with photon energy, which is most likely due to the use of a 2 mm resin on the detector. The overresponse was found to be of the order of a factor of 3 just above an effective photon energy of 50 keV.

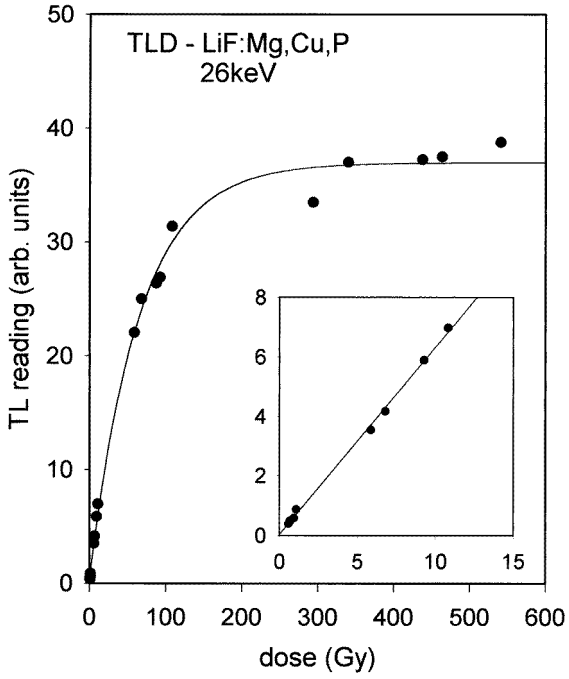

Figure 7. Dose response of LiF:Mg, Cu, P TL dosimeters. The full curve represents the best fit using the monoexponential function of equation (2). The inset shows the dose response up to 15 Gy magnified. Within the uncertainty of the measurements no supralinearity can be observed.

Table 3 lists the maximum dose response relative to water for all detectors included in the study. The data are normalized to the response to 6 MV spectral photons from a medical linear accelerator. The photon energy corresponding to the maximum response is also listed. All data except those for the radiochromic film were obtained using the fit of equation (3). For comparison data of Edwards *et al* (1997) are also included. These data were modelled from the results tabled in the original publication using the same equation (3).

3.5. Variation of the maximum response in different detector materials

From the theory of photon interaction with matter one can expect that the maximum of the dose response should increase with increasing effective atomic number of the detector material. At the same time the energy of the maximum response should increase as the photoelectric effect dominates the interaction up to higher photon energies. This is shown in figure 11, where the maximum dose response relative to water is plotted against the photon energy at which it occurs. The graph also includes data from Edwards *et al* (1997) which were derived from a fit of the data tabled in the publication using equation (3). For all detectors which do not include a specific build up or material which is aimed at improving the tissue equivalence of the detector the data can be well described using a monoexponential fit.

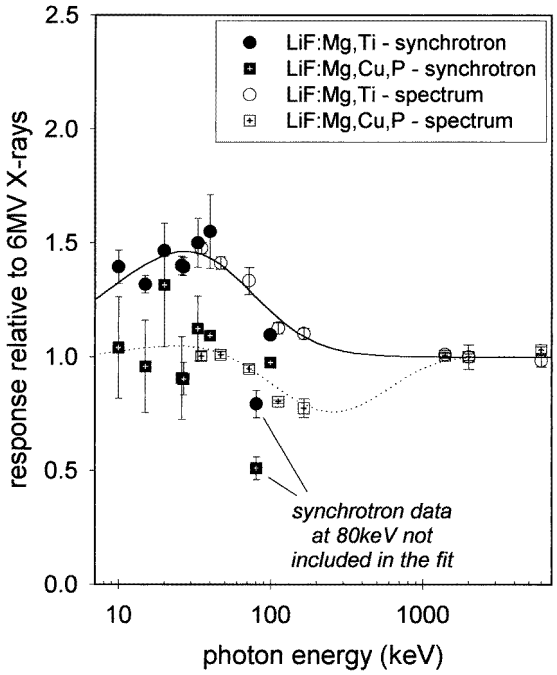

Figure 8. Variation of the dose response with effective radiation energy for thermoluminescence dosimeters (TLDs) relative to water. The response is normalized to 1 for 6 MV spectral x-rays from a medical linear accelerator. The curves represent the best fit according to equation (3). The values for 80.4 keV synchrotron radiation were not included in the fit.

4. Discussion

4.1. Dose response

If a detector does not exhibit a linear dose response there are different options to define dose response. Either a function with many parameters can be given or it can simply be linked to the response at a given dose or the dose required to achieve a certain level of detector response. In general the results will alter depending on what definition is used. In any case, the reliability of the results can be improved if an analytical model can be given which describes the variation of dose response.

The simple monoexponential fit used for the description of the dose response in both types of film appears to be adequate. It takes saturation of the detector response into account and assumes that background responses are corrected for. The monoexponential fit is a simplification of the equation of Williamson *et al* (1981) which also allows for variations of the optical density with depth in a phantom. Also the formula used by Mota *et al* (1990) to linearize the dose response for film after megavoltage photon irradiation is based on a monoexponential fit. The fact that a similar fit approximates low to medium (present study) as well as megavoltage photons supports the notion that dose response can be calculated similarly for all photon energies. This supports the use of a fixed detector response (for example the dose required to achieve a certain optical density) as a measure of dose response, as has been done in the present study. No complex model of dose response is required for the CEA film, as demonstrated by Cheng and Das (1996). These investigators

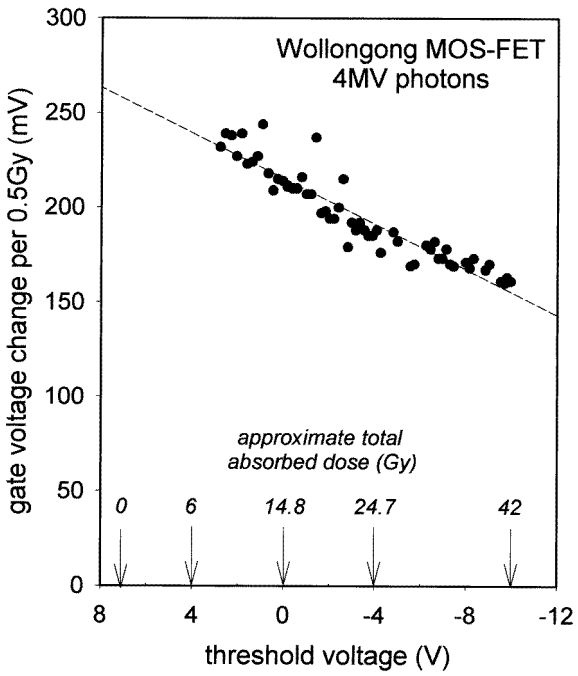

Figure 9. Variation of response with total absorbed dose in MOSFET detector A exposed to 4 MV photons from a medical linear accelerator (Varian Clinac 600C/4MV). The change required in gate voltage to compensate the charge accumulated due to 0.5 Gy absorbed dose was used as dose response. It is shown as a function of total absorbed dose and the actual MOSFET total gate voltage required to maintain a constant current between source and drain. The bias applied during irradiation was 5 V. The broken line shows the best linear fit to the experimental data ($r^2 = 0.877$).

found that the dose response for CEA TVS film was linear up to some 80 cGy with a slope of approximately one optical density unit per 20 cGy. This is similar to the dose response found in the present study.

The dose response in radiochromic film at 26 keV was also described adequately using a monoexponential fit. The optical density of 0.12 for 50 Gy delivered to MD-870 film is comparable with the results presented by Muench *et al* (1991) who found a response of just under 0.1 at an equivalent energy of 28 keV for the early GAF-chromic film (NA 7460-9). These investigators also studied the variation of curvature in the dose response curve with changes in the wavelength used for the densitometric evaluation. It seems to be appropriate to use the monoexponential fit of equation (2) to characterize each film/densitometer combination. Different degrees of curvature in the dose response curve have also been found by several other investigators who have studied various types of radiochromic film for megavoltage radiation (Meigooni *et al* 1996, Zhu *et al* 1997, Klassen and van der Zwan 1997).

The simple monoexponential model does not allow fitting of more complex dose response relationships like supralinearity of thermoluminescence dosimeters (Busoli 1981, Feist 1988). In this case either a more complex function has to be chosen or the fit has to be limited to the dose range of interest for a particular application (Feist 1988). In the case of LiF:Mg, Cu, P no supralinearity could be detected in the present study which is in

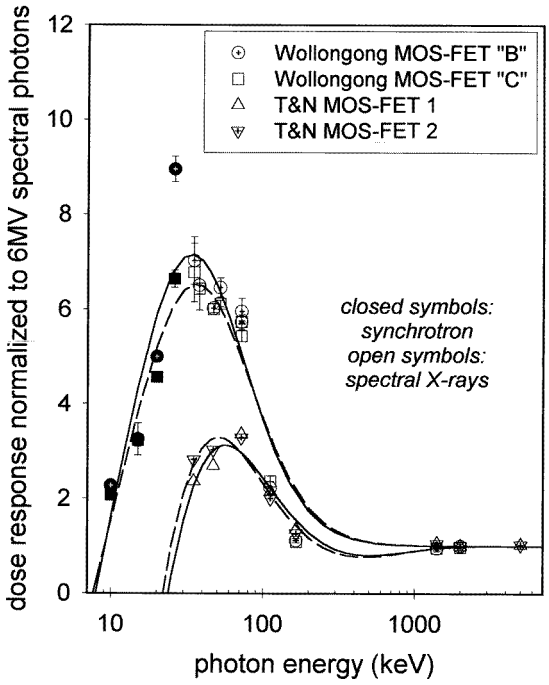

Figure 10. Variation of the dose response with effective radiation energy for MOSFET detectors. The response is normalized to 1 for 6 MV spectral x-rays from a medical linear accelerator. The curves represent the best fit according to equation (3).

agreement with the findings of Bartolotta *et al* (1995) who found a linear fit appropriate up to 10 Gy absorbed dose in LiF:Mg, Cu, P.

The relative dose response of both MOSFET detector types investigated in the present study was found to diminish with decreasing gate voltage, the latter being a measure of the total absorbed dose. This behaviour can be explained by positive charge trapping in the SiO_2 which leads to a reduction of the electrical field in the gate oxide. A similar reduction of response was observed by Soubra *et al* (1994) who found a reduction of dose response by 2.5% per absorbed Gy in single detectors. These researchers also found that the effect was more noticeable for lower bias voltages, which may explain in part the difference observed between the Thomson and Nielson detectors (bias 7 V) and the ones from the University of Wollongong (bias 5 V). The other contributing factor is the dual-bias dual-MOSFET design of the commercial system: the difference between two nonlinear dose response relationships is more linear than each of them.

For dosimetry in radiotherapy the nonlinearity of both MOSFET systems should be corrected for. This can easily be done, as can be seen in figure 9. Therefore, only the magnitude of the correction is different for the two detector types.

4.2. Variation of dose response with radiation quality

It is well known that radiographic film overresponds to low- and medium-energy photons compared with water, due to its high atomic number. Therefore most applications of film for dosimetric applications have been limited to electron dosimetry (Dutreix and Dutreix

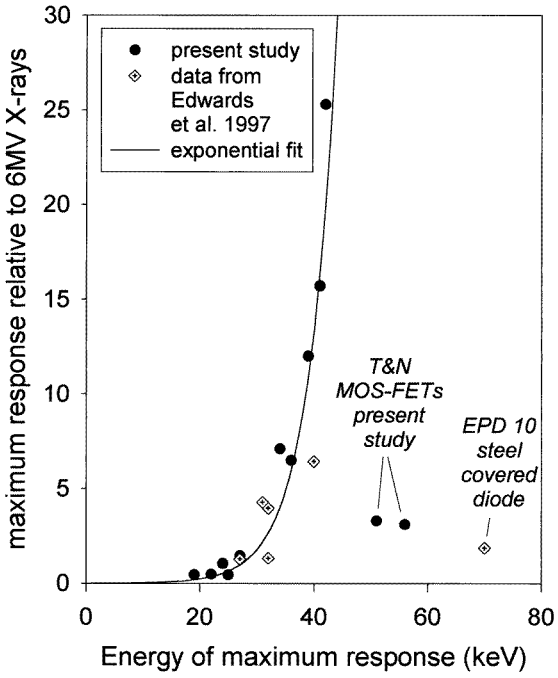

Figure 11. Maximum response to low- to medium-energy x-rays of a variety of radiation detectors. The maximum response is given as a function of the energy at which the maximum occurs.

1969, El-Khatib 1992). However, the availability of improved film types such as the CEA verification film with a linear dose response (Cheng and Das 1996) make film more attractive also for photon dosimetry. Similarly, the suitability of film for exit dose measurements from portal films (VanDam *et al* 1992, Ebert and Kron 1994) may stimulate new interest in photon film dosimetry. In these cases low-energy photons cannot be ignored, as demonstrated by Amols *et al* (1986). Therefore, the simple model described in equation (3) may prove useful to correct for the variation in dose response with photon energy. A suitable fit needs to be established for each type of film, as the difference found between Kodak X-Omat V and CEA TVS film shows. It also has to be established if processing conditions influence the dose response variation with x-ray energy.

To our knowledge there is no study published which describes the variation of dose response with photon energy for low- and medium-energy photons for CEA film. Cheng and Das (1996) note a difference of some 10% between the dose response to megavoltage x-rays from linear accelerators and ^{137}Cs radiation. This is roughly compatible with the results of the fit developed in the present study, where the overresponse of CEA film was found to be noticeable at photon energies around 500 keV. The difference between the maximum response for Kodak X-Omat V film and CEA film can easily be explained by the higher silver content in the latter. According to Cheng and Das (1996), CEA film has a silver concentration of 42 g m^{-2} as compared with 7 g m^{-2} in standard x-ray film. The higher silver concentration will increase the effective atomic number of the film, which in turn will result in a higher response to low-energy photons.

The variation of dose response with photon energy of radiochromic film was studied by Muench *et al* (1991) in order to provide correction factors for brachytherapy dosimetry.

He and his coworkers studied LiF:Mg, Ti and GAF chromic film. They found a TLD overresponse of 1.41 at 28 keV, which is in good agreement with the results of the present study. In GAF chromic film they only observed an underresponse of 0.7 as compared with less than 0.5 in the present study. This is due to the fact that Muench *et al* (1991) only investigated photon beams up to a minimum equivalent energy of 28 keV. As the response minimum in the present study was found at less than 25 keV, this would explain the difference.

The small reduction of dose response of LiF:Mg, Ti with increasing photon energy in the megavoltage range is comparable with the results published recently by Mobit *et al* (1996). These investigators studied the variation of dose response with x-ray energy in 1 mm thick LiF:MgTi samples and compared this with Monte Carlo calculation.

Edwards *et al* (1997) have described the variation of dose response with photon energy of MOSFETs. These investigators found an overresponse of slightly more than a factor of 4 which is in good agreement with the statement of the manufacturer (Edwards *et al* 1997). However, they found the energy of the maximum response to by approximately 32 keV which is some 25% lower than the manufacturer's statements.

Our results indicate a better tissue equivalence for the Thomson and Nielson detectors with a maximum of just above 3 at a photon energy of more than 50 keV. The low-energy photon response of the MOSFET dosimeters from Wollongong University showed qualitatively the same behaviour, but their response is higher by a factor of 2 than the Thomson and Nielson detectors. The lower relative dose enhancement for T&N MOSFETs can be explained by the sensor packaging. The 2 mm epoxy bulb on the top of the MOSFET chip acts as an additional filter for the low-energy part of the photon spectra. In addition to this, many secondary electrons registered in the detector will come from the surrounding medium because of the extremely small dimensions of the active detector volume. Therefore, the differences in the substrate for two MOSFET types may be significant: the Thomson and Nielson chip is placed on the plastic substrate whilst the Wollongong system utilized the MOSFET chip placed on the TO-5 gold-plated Kovar substrate. The difference in the number of backscattered electrons can explain some of the differences observed (Rosenfeld *et al* 1995).

As the energy response depends on these microscopic structures it may be difficult to reproduce the variation of dose response with photon energy from one detector to the next. This is illustrated by the differences in response found for both types of MOSFET detectors (compare table 3 and figure 10). In any case, the trade-off of the improved energy response is increased physical size, as evident in the Thomson and Nielson detectors.

In the literature the ratio of stopping powers or mass energy coefficients is often used to describe the variation of dose response with varying photon energy (Nariyama *et al* 1993, 1997). While this would be a good predictor of dose deposition in a homogeneous medium, it is only adequate to predict a trend in 'real' detectors. A variety of other factors influence the variation of dose response with photon energy. These include the mixture of different materials (such as in film and semiconductors) and the size of active grains (such as in radiographic film and thermoluminescence dosimeters). The latter is typically of the same order of magnitude as the range of secondary electrons of kilovoltage photons. Therefore, variation in grain size and density may influence where the secondary electrons are generated and where they deposit energy. This has been discussed for TLDs by Horowitz (1984) and Olko (1993).

Therefore, the energy response of a real detector may be difficult to predict from theory and must be assessed experimentally. In this case, a simple mathematical model to fit the observed variation in energy response is also useful. This has been attempted with the model

described in equation (3). As can be seen in figures 4, 8 and 10 the model successfully describes the variation of dose response with energy for a variety of detector types. The fits of data from the literature (Edwards *et al* 1997) proved to be good as well.

Unfortunately, the effective energy is neither an unambiguous nor a common description of spectral x-ray qualities. As such there may be circumstances where a fit of dose response variation with half value layer (HVL) of a beam may be more appropriate. This is illustrated in figure 12 for X-Omat film, LiF:Mg, Ti TLDs and the Wollongong MOSFET detectors. The HVL is expressed in mm Al and data provided by Seuntjens *et al* (1987) were used to convert the HVL measured using copper into HVL in mm of aluminium. The simple model of equation (3) still provides a reasonable fit of the experimental data. However, the physical reasoning of the decrease in photoelectric effect with energy as inverse cube does not apply.

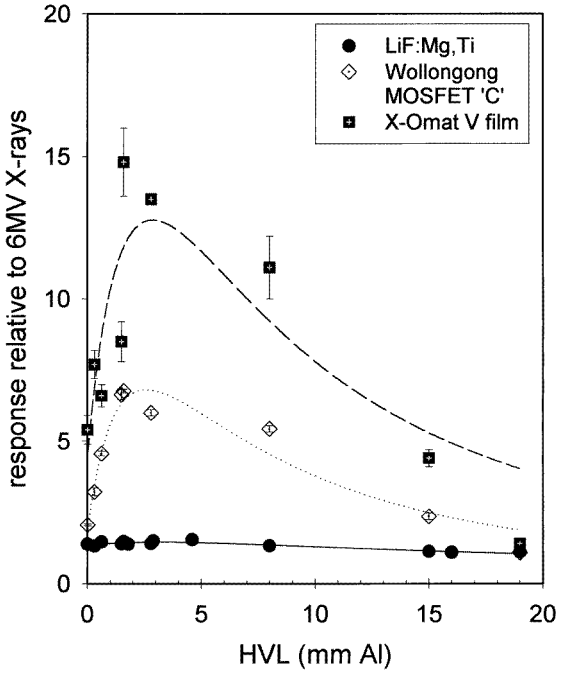

Figure 12. Variation of dose response of three detector types as a function of the half value layer (HVL) of the radiation beams. The response is given relative to the detector response to 6 MV x-rays from a linear accelerator. The curves show the best fit according to equation (3).

5. Conclusion

The results of the present study demonstrate the usefulness of monoenergetic photons for the study of the energy response of radiation detectors. All detectors studied would require a correction to determine the dose accurately in low- and medium-energy x-ray beams. The simple model developed to describe the variation of dose response with energy in the present study appears to be well suited to provide the basis for such a correction. However, even a precalculated correction may be difficult to apply as radiation conditions change. In addition to this, the differences observed in energy response for individual MOSFET

detectors and GAF chromic film emphasize the need for an individual correction if accurate dosimetry of low- to medium-energy x-rays is attempted.

Acknowledgments

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