The British Journal of Radiology, 72 (1999), 562–568

Evaluation of a MOSFET radiation sensor for the measurement of entrance surface dose in diagnostic radiology

D J PEET, BSc, MSc and M D PRYOR, BSc, MSc

Regional Radiation Protection Service, Royal Surrey County Hospital, Guildford, Surrey GU2 5XX, UK

Abstract. A patient dosimetry system using MOSFET technology (Thomson and Neilson Electronics Ltd, Canada) is evaluated for entrance surface dose measurements in diagnostic radiology. The system sensitivity for the standard MOSFET detector coupled to a high sensitivity bias supply was measured to be 1 mV mGy^{-1}. Response of a new high sensitivity dosemeter was measured to be 3 mV mGy^{-1}. The minimum detectable entrance surface dose at which a single measurement can be made with less than 25% total uncertainty at the 95% confidence level was estimated to be 4 mGy for the standard dosemeter and 1.5 mGy for the new high sensitivity dosemeter. The dosemeters were found to be linear with absorbed dose in air, linear with dose rate and reproducible, although they showed some energy dependence across the diagnostic energy range. The system is also compared with thermoluminescent dosimetry (TLD) as a tool for the measurement of entrance surface dose in diagnostic radiology. MOSFET detectors are considered to have advantages over TLD dosemeters with the instant readout of entrance surface dose. These dosemeters do have the disadvantage that they are visible in radiographs, they have a finite shelf life and can only accumulate absorbed dose up to a limiting value after which the dosemeters can no longer be used.

A patient dosimetry system using MOSFET technology has been developed by Thomson and Neilson Electronics Ltd, Canada as an alternative to thermoluminescent dosimetry (TLD) dosemeters for the measurement of patient dose in radiotherapy [1, 2]. The detector is a MOSFET device which when irradiated causes a permanent shift in the threshold voltage of the transistor. The threshold voltage is defined as the voltage required to allow current to flow between the source and the drain of a MOSFET device [1]. This shift is proportional to the absorbed radiation dose.

The semiconductor detector has an active depth of 1 μm making it ideal for skin dose measurements. The MOSFET detector is also very small (Figure 1), 0.2 mm × 0.2 mm active area, and can be read out instantly when connected to the reader. Up to five dosemeters can be connected to a bias supply simultaneously, allowing multiple readings to be made. The bias supply is then connected to a reader (Figure 2) after a radiation exposure to measure the threshold voltage shift in the detector.

This system was developed for radiotherapy use but is also available with a high sensitivity bias supply giving a reported increased response from 2.5 mV R^{-1} (0.3 mV mGy^{-1}) for the standard bias supply to 7.5 mV R^{-1} (0.9 mV mGy^{-1}) at diagnostic energies (Table 1). This increase in sensitivity has resulted in a suggestion that the system may be used in diagnostic radiology, particularly for skin dose measurements in interventional procedures [McKay, Private communication, 1996]. New high sensitivity detectors have subsequently been developed with a reported further increase in sensitivity of a factor of 3 when connected to the high voltage bias supply, giving a suggested response of 22.5 mV R^{-1} (2.6 mV mGy^{-1}). The sensitivity of the system can be increased by increasing either the positive gate voltage or the thickness of the oxide layer on the detector. The high sensitivity bias supply increases the positive gate voltage, whilst the new detectors have an increased oxide layer thickness. Sensitivity therefore increases with gate bias. The higher the electric field across the oxide, the more positively charged holes are trapped on the Si/SiO_2 interface, whilst the thicker the oxide, the greater the number of positive holes generated leading to a higher number trapped in the Si/SiO_2 interface. In these experiments, the gate bias was constant and the difference in sensitivities was due to the thickness of the oxide layer. The high sensitivity detectors had an oxide thickness which was twice that of the standard dosemeter.

In the UK, patient dose measurements are carried out following the National Dose Protocol [3]. For 10 common radiographic examinations, reference doses have been published ranging from

Received 17 August 1998 and in revised form 9 February 1999, accepted 24 February 1999.

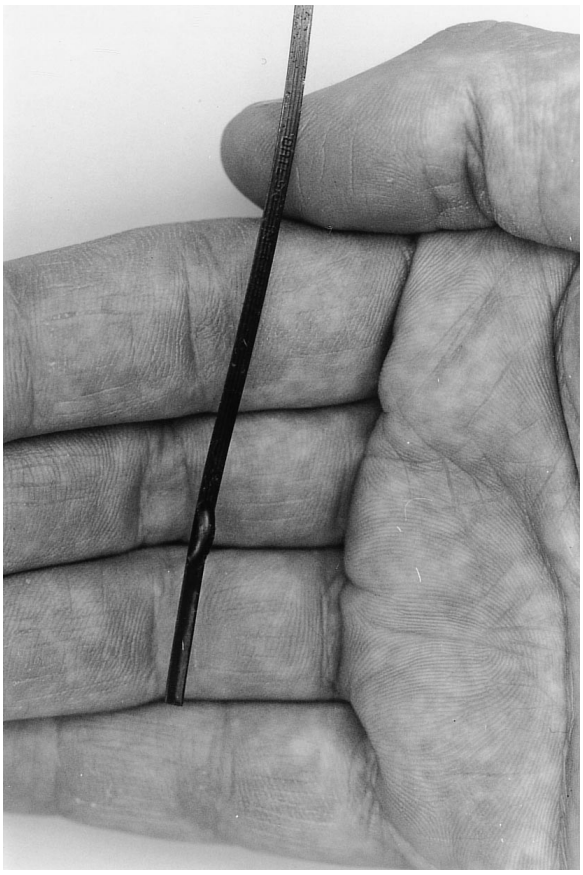

Figure 1. Photodetector on hand.

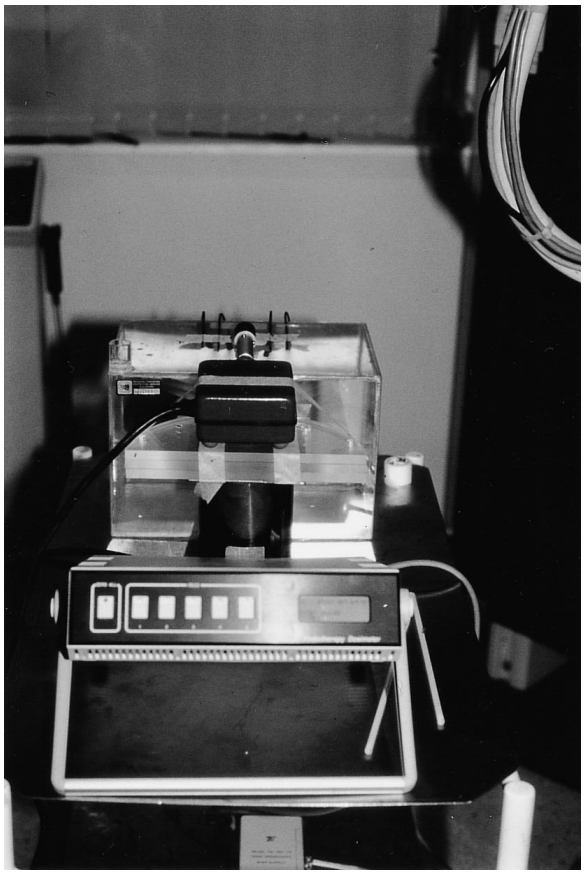

Figure 2. Photograph of calibration set-up/reader.

0.3 mGy for a posteroanterior (PA) chest to 40 mGy for a lumbar sacral joint (LSJ) view of a lateral lumbar spine with an anteroposterior (AP) abdomen being 10 mGy. These reference levels were set at the 75th percentile of entrance surface doses measured in a survey across the UK in 1982 [4]. Entrance surface dose is measured following the National Protocol on a sample of at least 10 patients weighing between 60 and 80 kg for the common examinations carried out in a particular hospital. The dosemeters used are usually TLD chips which have been calibrated so that any dose measured will have a total uncertainty of less than 25% at the 95% confidence level. The results obtained are compared with the reference levels and remedial action is taken in a centre where doses exceed national reference levels.

Since 1985, UK doses have been recorded in a national database and a review of these data was published in 1995 [5]. Because of the adoption of faster film–screen combinations across the UK, typically with a nominal speed of 400, doses were lower than those used to set the national reference levels. The entrance surface doses that 75% of centres were then achieving had dropped to 0.2 mGy for a PA chest and 36 mGy for the LSJ view. The third quartile value for the AP abdomen had dropped to 7 mGy.

At this centre, we routinely measure entrance surface dose for common radiographic examinations every 3 years following the methods in the National Dose Protocol. Since 1992, we have measured entrance surface doses on 4300 patients in 75 hospitals using lithium fluoride (LiF) TLD chips. The difficulties associated with the handling, annealing, batching and calibration of the TLD dosemeters, and the delay between the exposure of the patient and the readout of the entrance surface dose mean that the MOSFET system is potentially very attractive. The minimum entrance surface dose measurable is theoretically 0.4 mGy with the experimental dosemeters and high sensitivity bias supply. However, because of the uncertainties of measurement, the system cannot be used to give an entrance surface dose measurement of 0.4 mGy with less than 25% total uncertainty. At the start of this project it was recognized that the system was unlikely to be useful for PA chest measurements on patients but that it might be very useful for all other common radiographic views.

Table 1. Predicted sensitivities

Bias supply	Detector	Predicted sensitivity (mV mGy^{-1})
Standard	Standard	0.3
High sensitivity	Standard	0.9
High sensitivity	New high sensitivity	2.6

Measurements were made on the standard MOSFET connected to the high sensitivity bias supply and the new high sensitivity MOSFET connected to the high sensitivity bias supply to assess sensitivity, linearity, fading, energy response and to investigate whether the system could be used in the place of TLD for routine entrance surface dose measurements in diagnostic radiology following the UK National Protocol.

Method

All calibrations were made with the MOSFET detectors placed on a 20 cm water phantom to include full backscatter and measured against a Radcal 1015 dosemeter with a 6 cm^3 ion chamber attached. This dosemeter is a tertiary standard, calibrated at a UK NAMAS accredited laboratory. The equipment used for the irradiation of the detectors was a General Electric D38 mobile X-ray set with a single phase generator and an X-ray tube with 2.5 mmAl inherent filtration. The detectors were always attached to the same port on the high sensitivity bias supply box to remove any possible variations and orientated to have the build-up "bubble" facing away from the X-ray tube (into the water phantom). This was the recommendation from the manufacturer to give the highest sensitivity at diagnostic X-ray energies for the standard detectors.

The sensitivity of the detectors was measured at 80 kV, 200 mAs with a focus-to-skin distance (FSD) of 80 cm (Figure 2). After each exposure of the MOSFET detector and the ionization chamber, a reading in mVs was taken from the reader and the detector was zeroed and the reading from the ionization chamber recorded. After three exposures the detector and ionization chamber positions were reversed to take into account any variation in output across the field. This was repeated four times and an average calibration in mV per mGy was calculated for each detector.

The minimum entrance surface dose measurable with a single dosemeter was calculated by combining the uncertainty of readings of exposure from the Radcal 1015 of 4% at the 95% confidence level with the uncertainty of each reading on the Thomson and Neilson dosemeter of 1 mV at the 99% confidence level and 0.5 mV at the 95% confidence level.

Fading was examined by making an exposure every 15 min over a period of 2.25 h. The two standard sensitivity detectors were exposed and read out immediately. The sequence was then repeated with one of the detectors exposed but not read out until the end of the 2.25 h period.

Output linearity was assessed by making a series of 80 kV, 200 mA exposures for variations in time ranging from 0.1 to 1.6 s. Energy dependence of the new high sensitivity detectors was measured over a range 60–100 kV in 5 kV intervals using a Keithley 35080A kV divider to verify the set tube potential. A simple estimate of angular dependence was made by exposing the new high sensitivity detectors bubble side up and then bubble side down in air. Energy response was also measured in air over a range of tube potentials in the two orientations.

Having measured the calibration factors of the dosemeters under a range of exposure conditions, the appearance of the detector on a radiograph of a pelvic anthropomorphic phantom was assessed.

After discussion with radiologists and radiographers to ensure the phantom radiograph was acceptable, the appearance of the detector on clinical radiographs was assessed. An intercomparison between the MOSFET detectors and TLD material was also carried out. Two patients undergoing plain film abdomen (AP) examinations as controls for intravenous urography (IVU) examinations were asked to participate. A single high sensitivity detector was placed on the patient's midline, in the centre of the field of view. The bias supply was placed at the patient's side (Figure 3). Two sachets containing three TLD chips each were positioned either side of the detector. The exposure factors from each examination were recorded and the detector was read out and zeroed immediately.

The TLDs were calibrated using the same X-ray

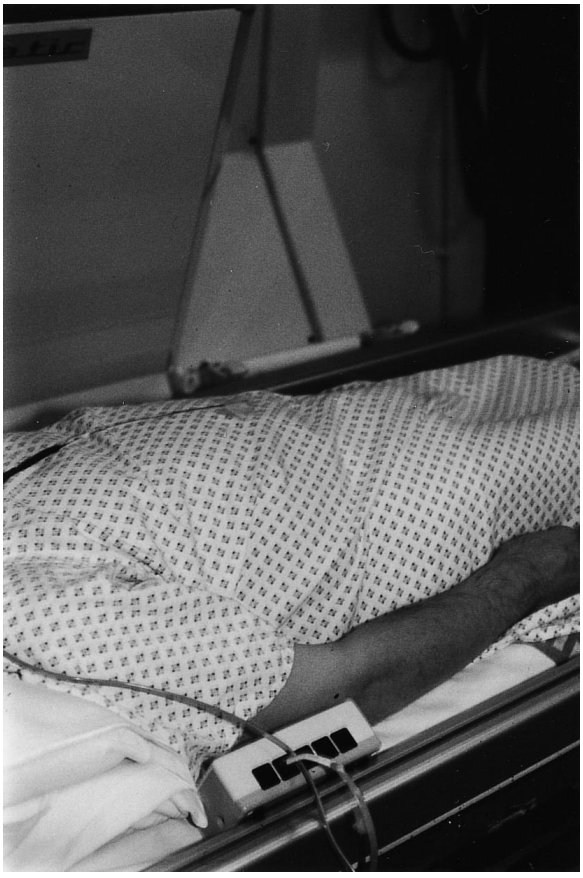

Figure 3. Photograph of patient.

Table 2. Measured sensitivities

Detector	mV mGy^{-1}
Standard 1	0.95
Standard 2	0.95
New high sensitivity 1	3.0
New high sensitivity 2	2.6

equipment and similar exposure settings. The sachets were positioned on a 20 cm water phantom with the Radcal 1015 ion chamber and the same MOSFET detector positioned either side. TLDs were read out using a Toledo reader and results compared with the MOSFET readings.

The films were viewed by a radiographer and a radiologist and opinions as to the visibility of the detectors and any diagnostic implications were noted.

Results

Four detectors were tested, two with standard sensitivity enhanced by the high sensitivity bias supply loaned by CIS (UK) and two new high sensitivity dosemeters loaned by Thomson and Neilsen. The new high sensitivity detectors were connected to the high sensitivity bias supply while measurements were being made on them. The sensitivity of each of the four detectors at 80 kV and a single dose rate in terms of mV per mGy are shown in Table 2.

The entrance surface doses measured using the standard and new high sensitivity dosemeters are shown in Table 3 with the total uncertainties in measurement at the 95% confidence level. These are calculated using the calibration factor of 2.6 mV mGy^{-1} for the new high sensitivity detector and 1 mV mGy^{-1} for the standard detector, and combine uncertainties in measurement of 0.5 mV for the MOSFET and 4% for the ionization chamber to obtain an uncertainty in the measurement of the calibration factor.

This is combined with the uncertainty at a range of readings to give an overall standard error. The total uncertainty is calculated as $2 \times$ the overall standard error divided by the measured entrance surface dose assuming that the overall error is a normal distribution.

The minimum detectable entrance surface dose with less than 25% uncertainty at the 95% confidence level for the standard dosemeter is just over 4 mGy and is 1.5 mGy for the new high sensitivity dosemeters.

Fading was not found to be a problem when exposed dosemeters were read out within 15 min. However, when two dosemeters were exposed every 15 min and one was read out immediately whilst the second dosemeter was read out at the end of the series of exposures, the readings on the second dosemeter were found to increase. The sensitivity of the dosemeter was 8.3 mV R^{-1} (1.0 mV mGy^{-1}) if exposed and read out within 15 min but this increased to 10.6 mV R^{-1} (1.2 mV mGy^{-1}) when the dosemeter was exposed every 15 min and read out after 2.25 h, causing an overestimation in exposure.

The linearity of two of the detectors, a standard detector and a new high sensitivity detector, is plotted in Figure 4 across a range of exposure times. The average calibration factor for the standard dosemeter was 0.9 mV mGy^{-1} and for the new high sensitivity dosemeter was 2.7 mV mGy^{-1}. The error bars are the total uncertainties including the random errors of the measurements.

The calibration factors measured at different tube potentials are plotted in Figure 5 for a standard MOSFET and the two new high sensitivity dosemeters.

The response in air and the marked angular dependence is shown in Table 4.

The detector can be clearly seen on the radiograph of the anthropomorphic phantom (Figure 6). However, the appearance of the detector on the films of patients is much less distinct

Table 3. Overall uncertainties

Dose (mGy)	% uncertainty at 95% confidence level	
	New high sensitivity detector	Standard detector
0.4	100	Below minimum sensitivity
0.8	50	Below minimum sensitivity
1.0	38	100
1.5	25	100
2.0	20	50
3.0	13	34
4	10	26
5	8	21
10	5	13
20	3.6	9
30	3.3	8

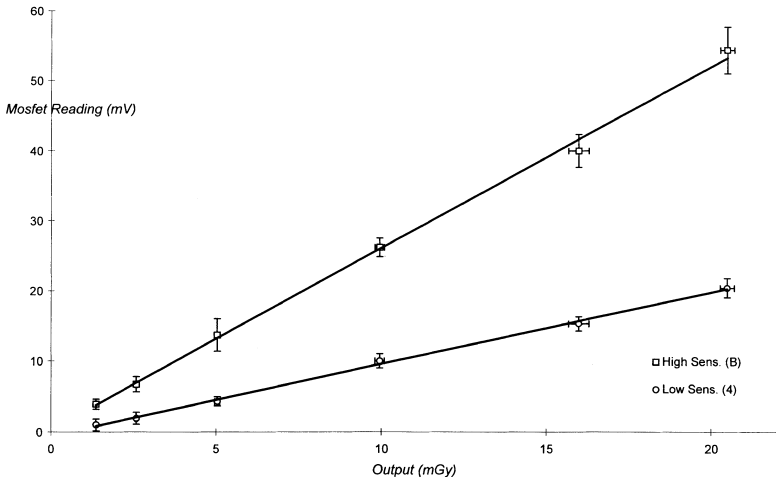

Figure 4. The linearity of a standard detector and a new high sensitivity detector plotted across a range of exposure times.

(Figure 7). The cable can just be seen to the left of the patient's midline (just to the right of the spinal column in the photograph).

The entrance surface doses measured for the two patients using TLD and the new high sensitivity MOSFETs were close and within the uncertainties of measurement. Measured entrance surface doses were very low compared with a typical AP abdomen, reflecting the use of an 800 speed film–screen combination in this room for this examination. The entrance surface dose measured for the first patient was 1.2 mGy using TLD and 1.7 using the MOSFET. Those measured for the second patient were 2.7 and 2.3 mGy, respectively, for the TLD and the MOSFET.

Discussion

The sensitivity of the standard detectors was measured to be 0.95 mV mGy^{-1} and that of the high sensitivity detectors 2.6–3 mV mGy^{-1}. The readings obtained were used to calculate the total uncertainties in each measurement system such that the standard dosemeter would be capable of measuring entrance surface doses down to 4 mGy and the high sensitivity dosemeters 1.5 mGy with less than 25% uncertainty at the 95% confidence level. This is the maximum level of uncertainty in a measurement of patient dose allowed in the UK National Dose Protocol [3]. Values lower than 4 mGy have been measured on 34% of AP abdomens measured at this centre between 1992 and the present, indicating that the sensitivity of the standard dosemeter is not high enough for routine measurements. The new high sensitivity dosemeters, however, can measure down to lower values and only 1% of our measurements have been below the 1.5 mGy threshold of this dosemeter.

The percentage of entrance surface dose measurements below the thresholds for the standard and the new high sensitivity dosemeters for the common examinations measured by this centre locally are shown in Table 5. The table also shows that this system cannot be used to measure the entrance surface dose for PA chests which comprise 25% of examinations in the UK [6], but the new high sensitivity dosemeters could be used to measure abdomens, pelvises and all views for lumbar spine examinations. The standard dosemeters could be used to measure lateral lumbar spines but lumbar spines comprise a relatively small proportion of examinations in the UK.

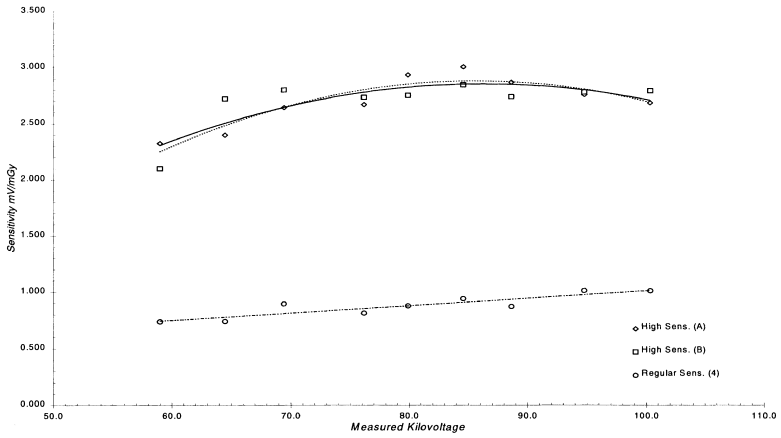

Figure 5. Calibration factors measured at different tube potentials for a standard MOSFET and the two new high sensitivity dosemeters.

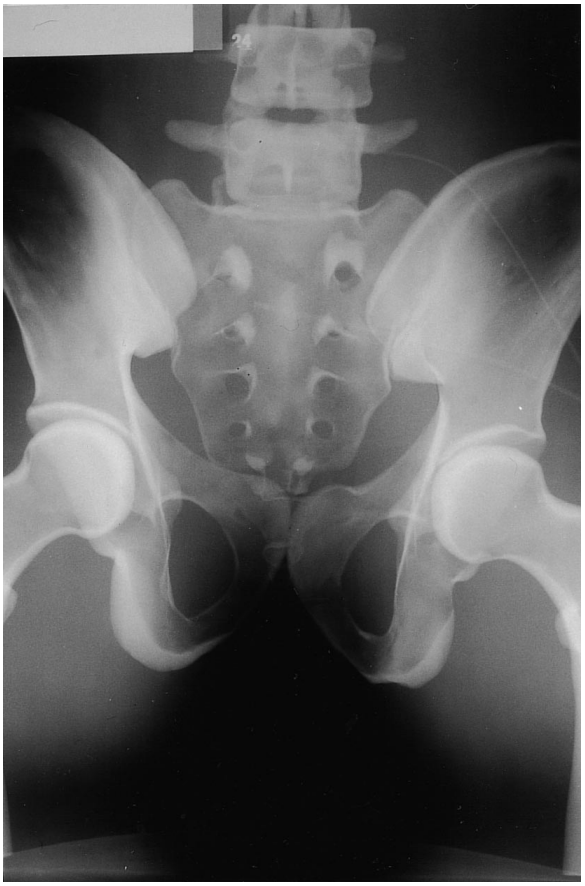

Figure 6. Phantom X-ray.

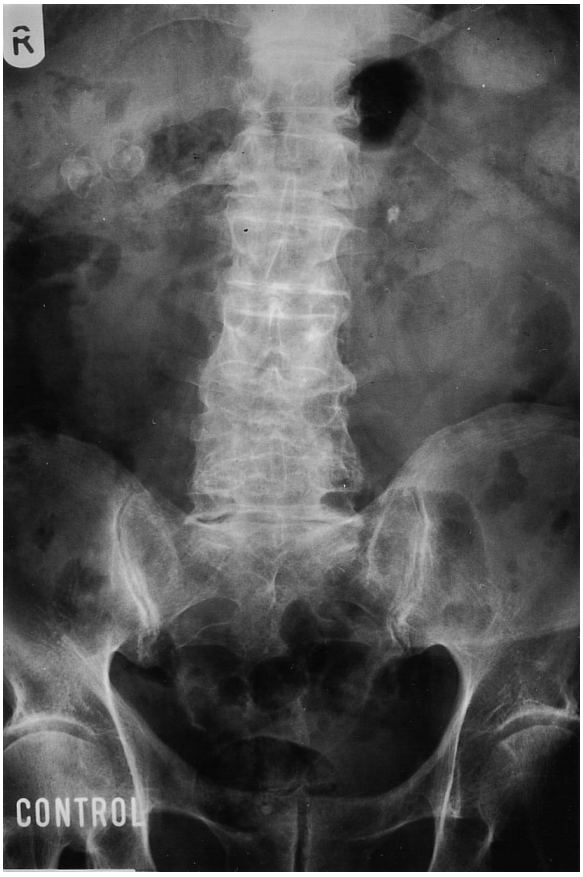

Figure 7. Patient radiograph.

Measurement uncertainties could be reduced by increasing the number of dosemeters used for each measurement, in the same way that more than one LiF chip is used for TLD measurements, although this was not assessed as part of this study.

Uncertainties could also be reduced by measuring a number of patients using the same dosimeter as is the case with our TLD measurements. However, we have confirmed the manufacturer's recommendation that the dosemeter should be read out within 15 min of each exposure. It would not be possible to measure more than one patient in most diagnostic radiology departments within this time scale. Fading for TLD dosemeters is less than 5% per year [7] using more than one dosemeter.

Linearity of the system is good over the range of exposures used diagnostically (Figure 4). The energy response is not identical over the range of energies used clinically (Figure 5). This is likely to be significant and would require a calibration of the dosemeter at each energy used for a particular examination.

Table 4. Angular dependence

Tube potential (kV)	Detector A (mV mGy^{-1}) (Bubble up)	Detector B (mV mGy^{-1}) (Bubble down)
60	3.8	2.2
80	3.3	2.6
100	3.7	2.5

The angular dependence (Table 3) is not important for measurements of entrance surface dose where the dosemeter can be placed precisely on a patient. The sensitivity measured in air with the bubble on the detector orientated upwards agrees well with the sensitivity reported elsewhere [8]. Our measurements gave between 3.3 and 3.8 mV mGy^{-1} (28.7–33.1 mV R^{-1}) compared with sensitivities of between 29.7 and 35.7 mV R^{-1} reported by Bower and Hintenlang [8].

The appearance of the dosemeter on the radiographs is likely to be considered important in some centres and not important in others. In this centre, the radiologist indicated that the cable resembled an ECG wire which they are quite used to seeing in radiographs although he suggested that the wire should be labelled. The TLD dosemeters are completely invisible on radiographs at diagnostic energies.

The detectors also have a finite lifetime of 6 months and a maximum accumulation of 20 000 mV or 7 Gy for the new high sensitivity dosemeters. This translates into approximately 1400 patients undergoing AP abdomen, pelvis or lumbar spine views although an allowance has to be made for the calibration of each dosemeter.

Table 5. Percentage of our measurements that could have been carried out using this dosimetry system

Examination	No. of patients	No. of measurements	% measurements <1.5 mGy	% measurements <4 mGy
Abdomen AP	502	86	1	34
Chest PA	1286	145	100	100
L. spine AP	630	100	0	33
L. spine lateral	643	102	0	1
L. spine L5/S1	195	34	0	3
Pelvis AP	453	89	0	46

The dosemeter is relatively expensive compared with other dosemeters used in diagnostic radiology although it costs much less than a TLD reader. The cost is comparable with the cost of the purchase and fitting of a dose–area product meter which is commonly used in the UK for the measurement of dose–area products in complex radiographic and fluoroscopic procedures.

No assessment was made during this project of the suitability of the dosemeter for the measurement of skin dose in fluoroscopy although the negative fading effect with time may become significant especially in interventional procedures which can have screening times of around 30 min and total examination times of up to 2 h. This is reported to be a complicated but reproducible function of time [8].

Conclusions

The MOSFET system is an alternative to the use of TLD dosemeters in the measurement of entrance surface dose for adult patients in diagnostic radiology with the exception of very low dose procedures such as PA chest. The system can be used to measure entrance surface dose on patients undergoing plain film radiography involving the abdomen, pelvis and lumbar spine regions using the new high sensitivity dosemeters. The standard dosemeters with the high sensitivity bias supply can be used to measure entrance surface dose for lateral lumbar spine projections but do not have sufficient sensitivity to measure the doses routinely received in AP abdomen pelvis and lumbar spine examinations in the UK today.

The system is linear and can be calibrated at the correct energy for clinical use. The immediate readout of entrance surface dose is the major advantage of this system and could be used by radiographers to audit doses and as a tool in dose reduction programmes for plain film radiography. The appearance of the dosemeter and cable on a radiograph is not considered to be a major disadvantage.

Acknowledgments

The authors are grateful to CIS (UK) for the loan of the standard detectors and dosemeter, Thomson and Nielsen for the loan of the new high sensitivity detectors and to the staff in the X-ray Department at the Royal Surrey County Hospital for their helpful comments.

References

1. Soubra M, Cygler J, Mackay GF. Evaluation of a dual bias dual metal-oxide-silicon semiconductor field effect transistor detector as a radiation dosimeter. Med Phys 1994;21:567–72.
2. Ramani R, Russell CMD, O'Brien P. Clinical dosimetry using MOSFETs. Int J Radiat Oncol Biol Phys 1997;37:959–64.
3. Institute of Physical Sciences in Medicine, National Radiological Protection Board, College of Radiographers. National Protocol for Patient Dose Measurements in Diagnostic Radiology. Chilton: NRPB, 1992.
4. Shrimpton PC, Wall BF, Jones BF, Fisher DG, Hillier MC, Kendall GM, et al. A national survey of doses to patients undergoing a selection of routine X-ray examinations in English hospitals, NRPB-R200. London: HMSO, 1986.
5. Hart D, Hillier MC, Wall BF, Shrimpton PC, Bungay D. Doses to patients from medical examinations in the UK 1995 review, NRPB-R289. Chilton: NRPB, 1996.
6. National Radiological Protection Board. Patient dose reduction in diagnostic radiology. Doc NRPB 1990;1: No. 3.
7. Marshall TO. Practical aspects of thermoluminescent dosimetry. In: Hufton AP, editor. HPA Report 43. York: IPSM, 1984.
8. Bower MW, Hintenlang HE. The characterisation of a commercial MOSFET dosimeter system for use in diagnostic X-ray. Health Phys 1998;75:197–204.