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● *Physics Contribution*

CLINICAL DOSIMETRY USING MOSFETS

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Purpose: The use of metal oxide-silicon field effect transistors (MOSFETs) as clinical dosimeters is demonstrated for a number of patients with targets at different clinical sites.
Methods and Materials: Commercially available MOSFETs were characterized for energy response, angular dependency of response, and effect of accumulated dose on sensitivity and some inherent properties of MOSFETs. The doses determined both by thermoluminescence dosimetry (TLD) and MOSFETs in clinical situation were evaluated and compared to expected doses determined by calculation.
Results: It was observed that a standard calibration of 0.01 Gy/mV gave MOSFET determined doses which agreed with expected doses to within 5% at the 95% confidence limit for photon beams from 6 to 25 MV and electron beams from 5 to 14 MeV. An energy-dependent variation in response of up to 28% was observed between two orientations of a MOSFET. The MOSFET doses compared very well with the doses estimated by TLDs, and the patients tolerated MOSFETs very well. A standard deviation of 3.9% between expected dose and MOSFET determined dose was observed, while for TLDs the standard deviation was 5.1%. The advantages and disadvantages of using MOSFETs for clinical dosimetry are discussed in detail.
Conclusion: It was concluded that MOSFETs can be used as clinical dosimeters and can be a good alternative to TLDs. However, they have limitations under certain clinical situations. © **1997 Elsevier Science Inc.**

MOSFET, Clinical dosimetry, TLD.

INTRODUCTION

Patient dose verification is an important part of quality assurance in treatment delivery, and one of the most commonly used techniques to verify dose is thermoluminescence dosimetry (TLD). This technique is popular because of the small size of the dosimeter, the ease of use on the patient, and the lack of a good alternative technique. The use of TLD for clinical dose evaluation has several limitations. For example, advance notice is required for sample preparation, frequent calibration is necessary, and TLD is quite sensitive to environmental conditions and handling procedures. Careful measurements have shown that it is possible to determine dose with a reproducibility of ±5% (95% confidence limit) (6). Kron *et al.* (5) showed TLD signal loss of up to 40% when using a contact, planchet-type heating instrument. This signal loss is dependent on the position of the TLD chip on the planchet. In another study, a 7% error was observed when using different trays in an automatic TLD reader (10). Thermoluminescence dosimeters with quoted specifications which matched to within ±5%, but from different manufacturers, showed up to a 40% variation in sensitivity (2). This is a major problem with the use of TLD for patient dosimetry. Also, obtaining moderate accuracy (5%) of dose determination with the routine use of TLD is quite tedious because of the complicated annealing, handling, and calibration procedures. To achieve greater precision and accuracy in such measurements, very stringent procedures are required, and these are often not practical for routine dosimetry. Further, the reproducibility of TLD measurements is not good. To achieve reasonable confidence in TLD result, several dosimeters must be irradiated and averaged, increasing the resources required for clinical dosimetry.

We have used metal oxide-silicon field effect transistors (MOSFETs)[1] as an alternate technique to TLD. The MOSFET is a sandwich-type device consisting of a P-type silicon semiconductor substrate separated from a metal gate by an insulating oxide layer. A negative bias applied to the gate causes a positive mirror charge to build up in silicon. This buildup of charge allows current to pass through the silicon substrate from the source to the drain terminal. The gate voltage necessary to allow conduction through the MOSFET is known as the threshold voltage. When the MOSFET is exposed to ionising radiation, elec-

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[1] Thompson and Nielsen Electronics Ltd., Ottawa, Canada.

tron hole pairs are formed in the oxide insulation layer. The applied positive potential to the gate causes the electrons to travel to the gate while the holes migrate to the oxide silicon interface, where they are trapped. These trapped positive charges cause a shift in the threshold voltage. The threshold voltage shift is proportional to the radiation dose deposited in the oxide layer. This is the basis of using MOSFETs as radiation dosimeters (3). The active region of these sensors is typically about 400 × 500 × 100 μm and the response is reported to be fairly insensitive to energy variations (1).

The use of MOSFETs as radiation detectors has been reported (3, 4), but their use in clinical dosimetry is limited as evidenced by the very few reports in the literature (1, 9, 11). Cygler *et al.* (1) reported the use of MOSFETs in clinical trials involving total body irradiation (TBI) and high-dose radiation (HDR) skin measurements. In TBI measurements, the reproducibility during patient dosimetry was shown to be comparable to diodes, and TLDs at the 4–5% level. For HDR exit dose measurements with a target breast dose of 5 Gy, the exit dose determined was comparable to TLD measured dose to within 6%. The application of MOSFET sensors for radiosurgery measurements was reported by Wu *et al.* (11). They measured the gamma knife helmet factor and compared it with factors measured by conventional methods using an ion chamber and TLDs. The percent differences reported were from −3.8% to 1.5%. A cone output measurement with Linac-based radiosurgery showed a difference in the range of −2.1–0.8%. All these results indicate the usefulness of MOSFETs as radiation sensors in therapy beams. The present report also deals with using MOSFETs in routine clinical dosimetry and includes measurements to demonstrate important physical characteristics of the sensors when used for clinical dosimetry.

METHODS AND MATERIALS

The dosimeter consists of four parts: (a) the MOSFET detector bonded with an epoxy to the end of a 20-cm-long, 2.5-mm-wide, and 0.4-mm-thick flexible cable, resulting in a flat surface on one side and a rounded, 1-mm-thick epoxy coating on the other; (b) A 9-V bias supply box; (c) a reader with a liquid crystal display; and (d) a printer, as shown in Fig. 1.

Each MOSFET dosimeter comprises two identical MOSFETs fabricated on a silicon chip, and each MOSFET operates at a different positive gate bias (9). The structure and operation of each MOSFET was described earlier. The measured difference between the threshold voltage shifts of the two sensors on irradiation is proportional to absorbed dose.

For dose measurements, the MOSFET sensors are plugged into the bias box, which is connected to the reader through an interface cable. The zeroing function measures and stores the total threshold voltage required to allow conduction through the MOSFET prior to irradiation. The

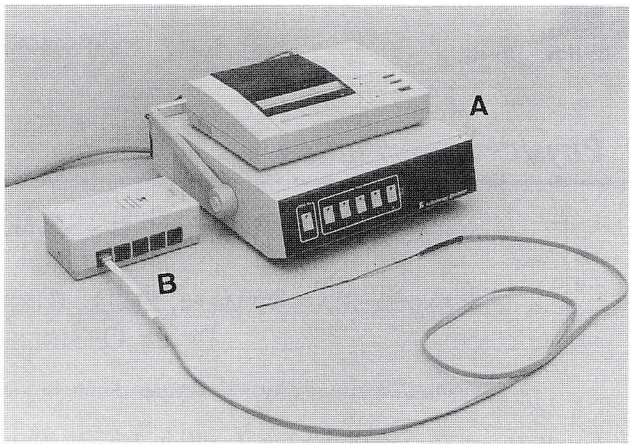

Fig. 1. Metal oxide-silicon field effect transistor (MOSFET) patient dose verification system consisting of (A) a MOSFET reader with a liquid crystal display display and printer positioned on top of the reader, and (B) a MOSFET sensor connected to a 9-V bias supply box.

threshold voltage shift, which is proportional to the radiation dose, is obtained by the read function following the irradiation. The MOSFET sensors were placed with flat side facing the beam and taped into position if required. The sensors must remain connected to the bias box (8 × 15 × 5 cm) during irradiation. After completion of the treatment, the bias box with MOSFET sensors is connected to the reader, and the dose received by the MOSFET is read by depressing the read key corresponding to the respective MOSFET sensor. The readings show the total threshold voltage in millivolts, the current threshold voltage difference in millivolts, and the corresponding dose in centigray. A hard copy of the data with the sensor identification can be printed.

For all calibration procedures, the MOSFET sensor was placed in a water phantom with the flat surface facing the beam at d_{max}, and an absolute dose was determined using an ionization chamber and the TG 21 protocol. A 3-min wait period was observed between each read operation performed on the reader. For each beam modality and energy, readings were taken for several doses over the range of 0.1–2 Gy and calibration factors were determined using linear regression and compared to manufacturer's preset value of 0.01 Gy/mV.

For estimating angular dependency, a spherical radiosurgery phantom as shown in Fig. 2 (7) was used. The MOSFET was placed in the center of an 8-cm-radius acrylic sphere with the flat side of the sensor facing the beam at 0° gantry angle. The center of the sphere was placed at the isocenter of a 6-MV photon beam treatment unit. Equally timed irradiations were performed at gantry angle intervals of 10° and the dose registered by the MOSFET was read from the reader for each beam. MOSFETs were also irradiated at 1.5 cm depth in a 6-MV photon beam both with the flat side and the curved, epoxy-coated surface facing the beam. Similar studies at 0° and 180°

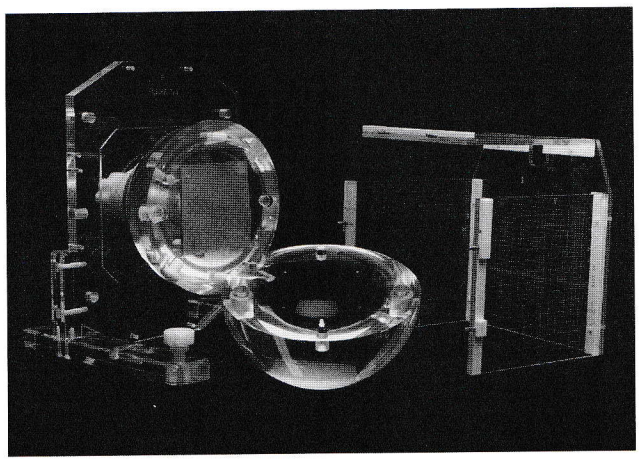

Fig. 2. Spherical acrylic radiosurgery phantom (7) used for angular dependency measurements.

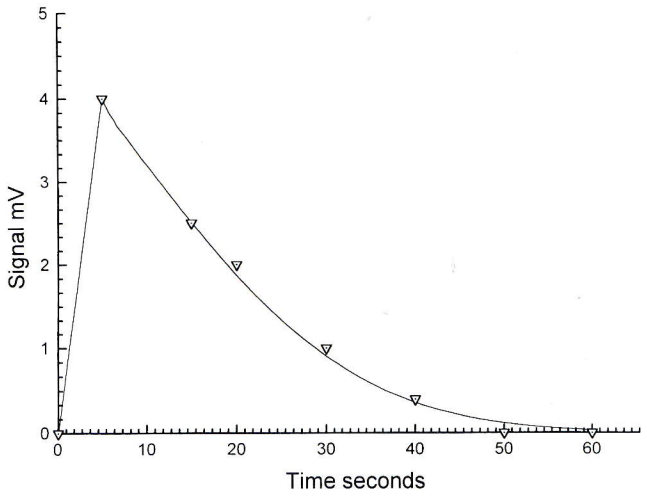

Fig. 3. The "creep-up effect" phenomenon.

gantry angles were carried out in ^{60}Co, 18-MV, and 25-MV photon beams. To study the accumulated dose effect, three MOSFETs were repeatedly irradiated to a known dose at d_{max} in a water phantom in a ^{60}Co beam. The MOSFET measured dose from these sensors was averaged. The average response (millivolts per centigray) of the MOSFETs was plotted as a function of accumulated dose.

Patient studies were done in two stages. Initially, to determine patient acceptance and establish MOSFET performance relative to our TLD system, the MOSFETs were used in parallel to our TLD system and the results were compared. Second, the MOSFETs were used alone on a series of patients and the results were compared to the expected result from calculation. In all cases, the factory preset calibration factor of 0.01 Gy/mV was used.

RESULTS AND DISCUSSION

Characterization of MOSFETs

The reproducibility of the readings from a MOSFET detector was measured by repeated exposures of 2 Gy. The standard deviation of the readings was 1%.

Table 1. Calibration factors for MOSFET in the range of 0.1–2 Gy

Energy	Calibration factor (Gy/mV)	Offset (mV)	Correlation coefficient
25-MV photon	0.0135	−3.03	0.9999
18-MV photon	0.01006	2.1	0.9998
6-MV photon	0.0126	−0.227	0.9999
^{60}Co photon	0.01015	−3.05	0.9999
19-MeV electron	0.010135	−3.03	0.9999
14-MeV electron	0.009928	2.49	0.9998
10-MeV electron	0.009934	3.35	0.9998
9-MeV electron	0.0101	2.256	0.9999
7-MeV electron	0.010106	−1.1033	0.9999
5-MeV electron	0.010246	0.6866	0.9999

A mean variation 0.0096 Gy with a standard deviation of 0.0056 Gy was observed between the MOSFET dose and the calculated dose, when irradiated in the range of 0.5–2 Gy.

The MOSFETs were calibrated using the procedure explained above in 6–25-MV photon and 5–19-MeV electron beams for a dose range of 0.1–2 Gy. The results are shown in Table 1, where the calibration factors are slopes (Gray per millivolts) obtained using linear regression and the offsets correspond to the intercepts of the straight lines (millivolts). The calibration factors for 6–25-MV photons and 5–19-MeV electrons range from 0.992 to 1.0135 with offsets from −3 to 3 mV. The variation in calibration factor over the entire range is approximately 3.3%. The average calibration factor is 0.01008 Gy/mV with an offset of 0.4 mV. The average calibration factor of 0.01008 should be compared to the factory preset calibration of 0.01 Gy/mV with an offset of 0.

The MOSFET is known to exhibit a phenomenon called "creep-up" (8), where the threshold voltage for the MOSFET increases with consecutive readings depending upon the time interval between successive read cycles. This effect occurs for accumulated doses of $\geq$ 20 Gy and is due to a charge being injected by the measuring circuit into the MOSFET, creating a temporary perturbation in the charge distribution. This perturbation normally decays in a few minutes if left unaltered, but it will be amplified if a second read pulse occurs before it has fully decayed. The resulting observation is a gradual elevation of the apparent dose. To evaluate this time decay phenomenon, we selected a MOSFET sensor which had received more than a 20-Gy dose. This sensor was then irradiated to a known dose. After the initial readout when a second readout was initiated at various time periods ranging from 0 to 1 min, the resulting measured dose in millivolts was recorded as a function of time. The results shown in Fig. 3 indicate that at the end of 1 min, the reading obtained was similar to the initial reading. The maximum increase observed

was 4 mV and was independent of the dose delivered to the MOSFET. The 4-mV increase corresponds to an apparent dose of 0.04 Gy. This dose is quite significant at low doses and for a typical clinical dose of 2 Gy will result in a 2% error. The readings taken beyond 1 min after the first read cycle show no "creep-up" effect.

The angular dependency from 0 to 360° of incident 6-MV radiation beam on to the MOSFET was determined using a spherical acrylic phantom with an 8-cm radius, as shown in Fig. 2. This ensures that the dose to the center of the phantom will be constant for any incident gantry angle. The MOSFET is placed at the centre of the phantom and oriented in such a way that the flat surface of the MOSFET faces the beam at 0° gantry angle, as shown in Fig. 4, where the long axis of the detector is oriented in and out of the plot. The measured doses per Monitor Unit (MU) at d_{max} after correcting for nontissue equivalence of the phantom are plotted in Fig. 4. These values are expected to be 0.01 Gy/MU. The angular effect was mainly seen between 140° and 220° gantry angles, where an overresponse of dose up to 15% was observed. Also, similar measurements were carried out with the gantry axis rotation perpendicular to the plane of plot in Fig. 4. The results were identical to the plot in Fig. 4. This difference in response between 0° and 180° gantry angles was reproducible and must be included in clinical dose estimations. The MOSFET responses at 0° and 180° gantry angles were

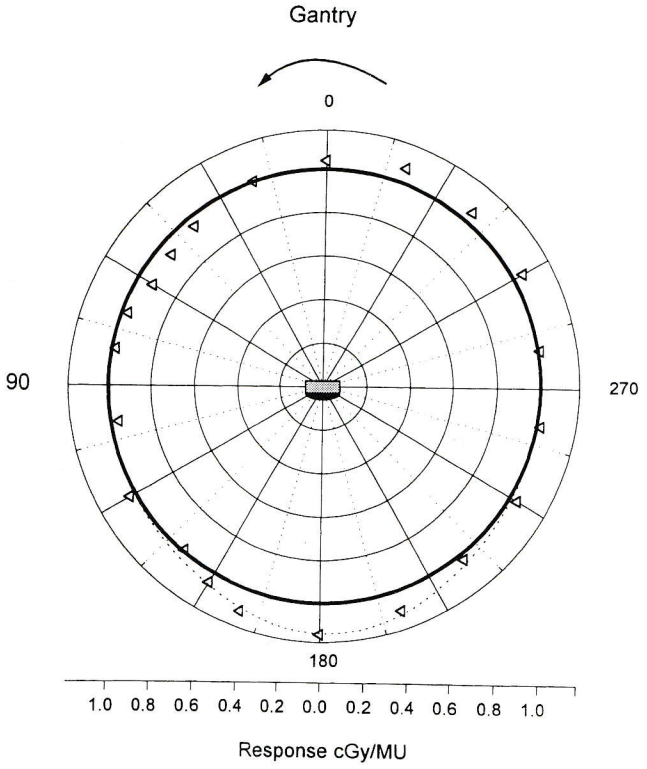

Fig. 4. Angular dependency from 0° to 360° of incident 6-MV photon beam onto the metal oxide-silicon field effect transistor (each circle has a radius of 0.002 Gy/MU response).

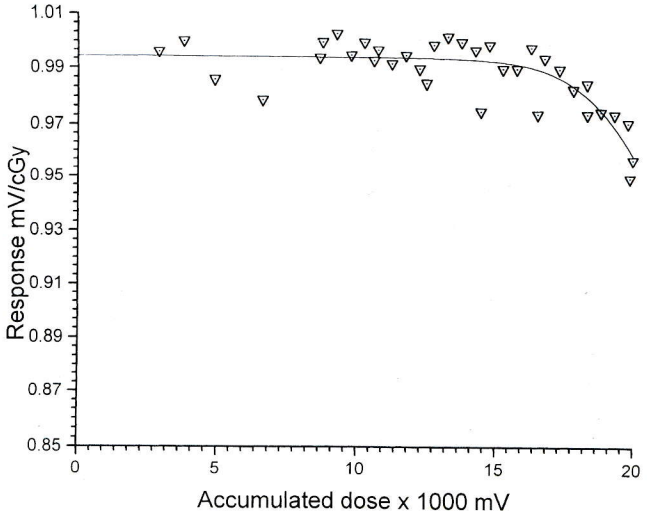

Fig. 5. Effect of accumulated dose on the relative response of the metal oxide-silicon field effect transistor sensor.

measured for ^{60}Co, 6-MV, 18-MV, and 25-MV photon beams at d_{max}. The observed overresponses at 180° gantry positions were 28%, 18%, 13%, and 11%, respectively. A detailed study on the angular effect of irradiation on MOSFETs is under way at our center.

The effect of accumulated dose on the relative response of the MOSFET was investigated, and the result is shown in Fig. 5. The response of the sensor (millivolts per antigray) decreased very marginally with accumulated dose, except toward the tail end of the recommended range, where a drop in response was observed. The overall variation in the response was about 2% between 0 and 180 Gy. Hence, we propose to limit the use of MOSFET to 180 Gy of the accumulated dose.

Our preliminary measurements on temperature dependence from room temperature to 37°C show an effect of 2–3%, which is not very significant in clinical dosimetry. Also, the MOSFET is placed directly on the patient body for dose verification, with the epoxy side in contact with the patient. The epoxy acts as a good insulator and prevents MOSFET sensor from being heated by the body temperature.

Clinical studies

In vivo dosimetry is usually limited to the patient surface. Calculation of expected doses at the patient surface can be very difficult because of dose build-up considerations, scattered radiation from shielding and treatment unit components, and nonuniformity in patient contour. Our patient cohort had a large proportion of electron fields owing to our experience with TLDs on these beams, where dose measurements often differed from expected results by up to 20%. We have used TLDs as a cross reference to compare the patient dose determined by MOSFET. Two to four TLDs and two MOSFETs were placed side by side in the field. The dose measured with MOSFETs and TLDs

Table 2. Comparative patient dose measurements between MOSFET and TLD

Site	Beam quality	MOSFET dose (Gy)	TLD dose (Gy)	Expected dose (Gy)
Breast	9-MeV electrons	1.6	1.65	1.64
Anterior neck	12-MeV electron	2.793	3.056	2.78
Chest wall boost	10-MeV electron	1.587	1.675	1.70
Direct back	10-MeV electron	2.97	2.893	3.00
Anterior nose	19-MeV electron	3.335	3.385	3.16

were averaged for each site and are reported in Table 2, along with the expected dose.

The MOSFET measured dose agreed with the expected dose to within 0.8% with a $\sigma = 3.9\%$, while for TLD the agreement is within 2.5% with a $\sigma = 5.1\%$. In general, the doses measured by MOSFETs and TLDs are in good agreement.

Having established agreement between TLD and MOSFET determined doses, the MOSFETs were evaluated as a standalone system. The results are listed in Table 3. The mean variation from the expected dose in this instance was 2.3% with a standard deviation of 4.1%. Once again, the measured dose compares very well with the expected dose.

To evaluate patient acceptance of MOSFET, each patient was interviewed regarding his comfort and the noticeability of the detector or wires. All but one of the patients interviewed did not notice the detectors, did not feel uncomfortable, and did not notice the wires. One patient refused the use of MOSFETs for dose determination, since the system was under investigation. The use of the detector did not produce any visible skin reaction or irritation. Our experience showed that the wire was flexible enough to position at various sites. The detector also fitted well under the 2-mm acrylic masks used at our center for immobilization of head and neck patients.

Drawbacks

1. The angular dependency of response between 140° and 220° beam entry positions onto MOSFETs is quite significant in clinical dosimetry. This result was highly reproducible and would be a problem for integrated measurements in AP/PA opposed beams, where the dose could be overestimated.
2. Toward the end of the recommended lifespan of the sensors (20,000 mV), at about 18,500 mV, the MOSFET sensor response suddenly decreases by about 5%. It is best not to use the sensor beyond 18,000 mV of the accumulated dose. It is therefore advisable to keep track of the total accumulated dose for each sensor, to ensure that there is enough remaining range for the dose measurements under consideration. This should be relatively easy to achieve, as for each zeroing operation performed the total accumulated dose in millivolts is printed out.
3. The "creep-up" effect discussed earlier could have a significant effect if a reading is initiated within a minute of previous read function, especially during calibration procedures. A wait period of at least 3 min is recommended before the next read function is initiated.

Table 3. Clinical dosimetry with MOSFETs

Site	Beam quality	MOSFET dose (Gy)	Expected dose (Gy)
Chest wall	13-MeV electron	2.448	2.40
Left lateral ear	6-MV photon + 19-MeV electron	1.3	1.344
Direct left ear	7-MeV electron	3.945	4.0
Direct left ear	10-MeV electron	2.755	2.5
Right cheek + neck	13-MeV electron	3.117	3.0
Direct right breast	10-MeV electron	1.795	1.705
Nostril	7-MeV electron	4.13	4.22
Right breast	10-MeV electron	1.64	1.705
Pelvis	T25-MV photon	2.075	2.0
Right eye	6-MV photon	0.17	0.21
Right eye	6-MV photon	0.145	0.131

CONCLUSIONS

The use of MOSFET for clinical dosimetry has several advantages. The simple calibration procedures, immediate readout, and ease of maintenance and operation of the equipment are some of the advantages of using this technique. Significantly less time is required for system maintenance, since annealing is not required. The dose accuracy and repeatability are not significantly different from TLD measurements, and individual MOSFETs have a higher reproducibility of response than individual TLDs. The advantage of the relatively small size of TLDs is offset by the tedious procedures required to use them. TLDs are highly susceptible to impurities and are sensitive to environmental changes, and hence require packaging to be used on the patient. MOSFETs are comparably sized to TLDs but do not have any dependency on impurities or environmental conditions and do not require packaging. The use of MOSFET is restricted by the present length of the cable, but the cables are small and flexible, and accepted by the patients. Some of the drawbacks for using MOSFETs discussed above could pose a problem for special techniques or sites, such as AP/PA radiation or dose to eye lens. In conclusion, it can be said that MOSFET dosimetry is a cost-effective, easy to use technique, and serves as a good alternative to TLD, with certain limitations.

REFERENCES

1. Cygler, J.; Soubra, M.; Szanto, J.; Mackay, G. F.; Thompson, I. Application of MOSFET dosimetry system in TBI and HDR treatments. Med. Phys. 22:1541; 1995 (abstr).
2. Fairbanks, E. J.; DeWerd, L. A. Thermoluminescent characteristics of LiF:Mg, Ti from three manufacturers. Med. Phys. 20:729–731; 1993.
3. Gladstone, D. J.; Chin, L. M. Automated data collection and analysis system for MOSFET radiation detectors. Med. Phys. 18:542–548; 1991.
4. Huges, R. C.; Huffman, D.; Snelling, J. V.; Zipperian, J. E.; Ricco, A. J.; Kelsey, A. Miniature radiation dosimeter for *in vivo* radiation measurement. Int. J. Radiat. Oncol. Biol. Phys. 14:963–967; 1988.
5. Kron, T.; Buston, M.; Wong, T.; Metcalfe, P. Readout thermoluminescence dosimetry chips using contact planchet heater. Australas. Phys. Eng. Sci. Med. 16:137–142; 1993.
6. McGhee, P.; Humphreys, S.; Dunscombe, P. An efficient approach to routine TL dosimetry. Med. Dosim. 18:187–191; 1993.
7. Ramani, R.; Ketko, M. G.; O'Brien, P. F.; Schwartz, M. L. A QA phantom for dynamic stereotactic radiosurgery: Quantitative measurements. Med. Phys. 22; 1343–1346; 1995.
8. Spender, M. Private communication. Thomson and Nielsen Electronics, Ottawa, Canada.
9. Soubra, M.; Cygler, J.; Mackay, G. Evaluation of dual bias dual metal oxide-silicon semiconductor field effect transistor detector as radiation dosimeter. Med. Phys. 21:567–572; 1994.
10. Wood, J. J.; Mayles, W. P. Factors affecting the precision of TLD dose measurements using an automatic TLD reader. Phys. Med. Biol. 40:309–313; 1995.
11. Wu, A.; Maitz, A.; Shortt, K.; Mackay, G.; Thompson, I.; Johnson, M.; Washick, J.; Lunsford, L.; Kalnicki, S. Performance of a new dosimetry system with MOSFET sensor for radiosurgery applications. In: AAPM, Anaheim, CA, July 25–28, 1994.