Comparison of peripheral dose measurements using Ionization chamber and MOSFET detector

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ABSTRACT

BACKGROUND: In radiation therapy, the peripheral dose (PD) – the dose outside the geometric boundaries of the radiation field – is of clinical importance. A metal oxide semiconductor field effect transistor (MOSFET) detector is used to estimate the peripheral dose.

AIM: The aim of this study is to investigate the ability of a MOSFET dosimetry system to accurately measure doses in peripheral regions of high energy X-ray beams.

MATERIALS & METHODS: The accuracy of the MOSFET system is evaluated by comparing peripheral region dose measurement with the results of standard ionization chamber measurements. Furthermore, the measurement of PD using a MOSFET detector helps us to keep the tolerance dose of any critical organ closer to the treatment field within the acceptable limits. The measurements were carried out using a 0.6 cc Farmer type ionization chamber and MOSFET 20 dosimetry system for field sizes ranging from 5 x 5 cm^2 to 20 x 20 cm^2 at three depths of 1.5 cm, 5 cm and 10 cm in a blue water phantom. PD were measured at distances varying from 1 cm to 30 cm from the field edges along the x axis for the open fields, with collimator rotation and with beam modifiers like 15 degree, 30 degree and 45 degree wedges.

RESULTS: The results show a good agreement of measured dose by both methods for various field sizes, collimator rotation and wedges.

CONCLUSION: The MOSFET detector has a compact construction, provides instant readout, is of minimal weight and can be used on any surface.

KEY WORDS: peripheral dose, MOSFET, radiation dosimetry

BACKGROUND

Radiation doses to critical organs outside the primary radiation field usually must be evaluated prior to treatment and reduced when necessary. In radiation therapy, the peripheral dose (PD), or the dose outside the geometric boundaries of the radiation field, is of clinical importance when surrounding normal structures with low dose tolerances are involved. Such structures include the gonads, the lenses of the eyes, the contralateral breast during breast treatment, and the fetus in a pregnant patient. Even a small percentage of the total prescribed dose might cause damage to critical structures.

The American Association of Physicists in Medicine Task Group 36 (AAPM TG-36) data can be used to estimate peripheral dose (PD) distributions for various treatments and to determine the need for additional shielding [1]. Peripheral dose is due to three main sources: 1. leakage from the treatment unit; 2. scatter from the secondary collimators and from beam modifiers such as wedges and blocks; and 3. internal scatter originating in the patient. The first two sources depend on the configuration of the treatment unit head, and therefore might be affected by changes in the design of the head and or additional beam

modifiers placed in the path of the beam. Hence the PD distributions consist of internal scatter, collimator scatter, and transmission through collimation, head leakage, and room scatter. The use of a tertiary MLC has been shown to significantly reduce PD due to reduction in scatter from the primary and secondary collimator, transmission through the secondary collimator, and head leakage [2]. It was also shown that intensity-modulated radiotherapy (IMRT) with increased monitor units (MU) ranging from a factor of 2 to 10 depending on the IMRT technique, compared to conventional therapy, led to increased PD [3]. The influence on PD by internal scatter, head scatter and head leakage varies with distance and dominance in contributing to PD. In vivo dosimetry for radiotherapy patients often requires dose measurements not in the treatment area, but in the peripheral regions, so that doses to critical organs can be recorded and minimized, if possible. For many such measurements, one needs detectors with the ability to measure low doses accurately and tolerance to some variations in the spectral quality of the calibration beam.

Detectors such as radiochromic film were not sensitive enough to measure small doses accurately [4]. The metal oxide semiconductor field effect transistor (MOSFET) featured the ability to integrate dose measurements and to provide immediate dose readout [5].

AIM

In combination with a very small sensing volume, it makes the MOSFET dosimetry system advantageous over the other systems used in radiotherapy. Thus, a MOSFET detector finds applications in radiotherapeutic in vivo dosimetry. We report the results of an investigation of the ability of a MOSFET dosimetry system to accurately measure doses in peripheral regions of high-energy X-ray beams; the accuracy of the MOSFET system is evaluated by comparing its peripheral dose with that of standard ionization chamber measurements.

METHODS & MATERIALS

Slab phantom measurements with ion chamber:

PD measurements were performed on a 6 MV linear accelerator (Siemens Primus) equipped with a multileaf collimator (MLC). The MLC replaces the lower secondary collimator on a conventional linear accelerator. The MLC consists of 26 leaf pairs, each leaf projection having 10 mm leaf width at the isocentre. All the measurements were carried out using a 0.6 cc Farmer type ionization chamber (PTW, Friedberg, Germany) inserted at 1.5 cm, 5 cm and 10 cm into a blue water phantom (Standard Imaging Inc.).

Each slab set was configured of blue water slab with area of 30 x 30 cm^2 placed on top of another to give 20 cm thickness. Three sets were lined up longitudinally to compose a phantom of 90 cm length, 30 cm width, and 20 cm depth.

By examining the effective field sizes in clinical practice we found that effective field sizes ranging from 5 x 5 cm^2 to 20 x 20 cm^2 were suitable to perform peripheral dose measurements. We therefore measured PD for 5 x 5 cm^2, 10 x 10 cm^2 and 20 x 20 cm^2 field sizes at three depths of 1.5 cm, 5 cm and 10 cm. The charge recorded by the electrometer in the picogray (pGy) range was measured for the above field sizes and depths.

Source to surface distance (SSD) was kept fixed at 100 cm. A dose of 100 monitor units (MU) was delivered to the detector at the central axis and the PD was measured at distances of 1 cm, 2 cm, 3 cm, 4 cm, 5 cm, 10 cm, 15 cm, 20 cm, 25 cm and 30 cm from the field edges along the x axis. Since the detector distances were measured from the geometric field edges at depth, the distances from the central axis to the detector vary with depth and field sizes.

PD was calculated as the percentage of dose on the central axis for each field size and depth by taking the ratio of the readings for the detector at the stated distance from the field edge to the readout of the detector on the central axis. Collimation was altered to isolate the influence of the collimation system and or orientation. Therefore, we measured data in the peripheral dose plane as collimated by the collimating jaws alone, with collimator 90° for field size 10 x 10 cm^2.

To evaluate the variation in PD while using wedges, the same measurements were repeated for 15° wedge, 30° wedge and 45° wedge for field size 10 x 10 cm^2. In this case

the measurement was made in such a way that the detector position was moving away from the thick end of the wedge.

Slab phantom measurements with MOSFET detector:

The metal oxide semiconductor field effect transistor (MOSFET) supplied by Thomson and Nielson is also evaluated for the estimation of peripheral dose measurement. The MOSFET 20 patient dose verification system with model TNRD-50 was used for this study. The dosimeter is composed of a 1.4 metre cable attached to a length of thin, semi-opaque polyimide laminate. The silicon detector itself was mounted on the end of the polyimide material under a 1 mm layer of black epoxy. The dual bias supply provides a choice of two sensitivities in order to cover a wide range of doses with optimum reproducibility. This bias supply allows you to choose between two options, i.e. standard or high sensitivity. Calibration was performed prior to the first measurement with a new dosimeter and periodically during the dosimeter's lifetime, using doses previously characterized with an ion chamber. Calibration should be performed under equilibrium conditions. It was usually done with full build-up material. For this study we selected the high sensitivity detector TN1002RD in combination with high bias supply which will give 9mV/cGy sensitivity for cobalt 60 gamma rays energy [6]. Two detectors were pasted onto the blue water phantom. Grooves were made in the blue water phantom to accommodate the MOSFET detector. A dose of 100 MU was delivered to the MOSFET detector at the central axis and doses in the peripheral region of the beam ranging from 1 cm to 30 cm from the field edge were measured as done with the ionization chamber. The doses measured with MOSFET were then analyzed with the results of the cylindrical thimble type ionization chamber.

RESULTS

The measured PD data for 6 MV X-rays at 1.5 cm, 5 cm and 10 cm depths as a function of distance up to 30 cm from the field edge for both the ionization chamber and the MOSFET detector are presented. The dose values were normalized independently to 100% for each

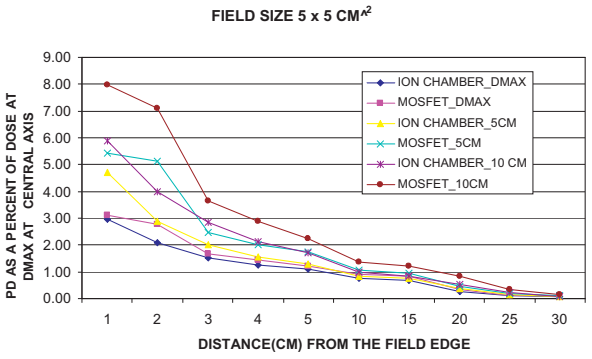

Fig. 1. Peripheral dose (PD) in phantom for 6 MV X-rays measured at dmax, 5 cm and 10 cm depth using ionization chamber and MOSFET detector for field size 5 x 5 cm^2

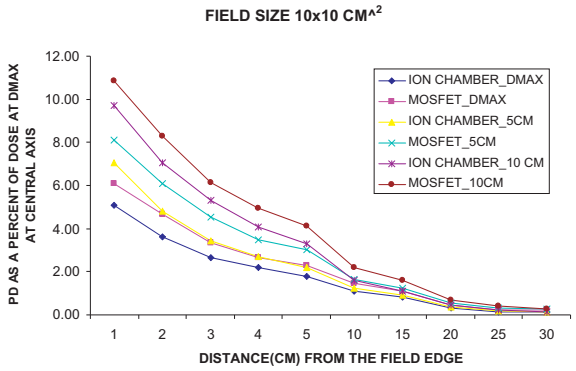

Fig. 2. Peripheral dose (PD) in phantom for 6 MV X-rays measured at dmax, 5 cm and 10 cm depth using ionization chamber and MOSFET detector for field size 10 x 10 cm^2

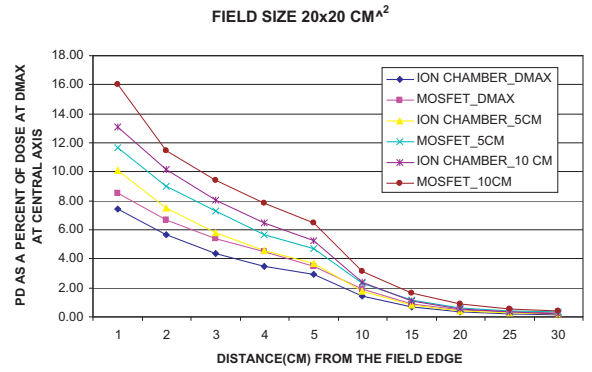

Fig. 3. Peripheral dose (PD) in phantom for 6 MV X-rays measured at dmax, 5 cm and 10 cm depth using ionization chamber and MOSFET detector for field size 20 x 20 cm^2

field size and depth on the central axis. Figures 1, 2, and 3 show the measured peripheral doses for the MOSFET detector and ionization chamber for 5 x 5 cm^2, 10 x 10 cm^2 and 20 x 20 cm^2 respectively. The results show a good agreement of measured dose by both methods with various field sizes. From the graphs it is clear that as the field size increases there is an increase in PD. It is also seen that at higher depths the PD increases for a given field size. We also found that the PD measured with MOSFET is slightly higher than that measured with the ionization chamber.

The measured PD data for 6 MV X-ray with field size of 10 x 10 cm^2 at 10 cm depth for 0° collimation and 90° collimation are shown in Figure 4. Collimator rotation results in a relatively small difference in PD distributions, and there does not seem to be a clear advantage of positioning the collimator to a certain setting to reduce the PD.

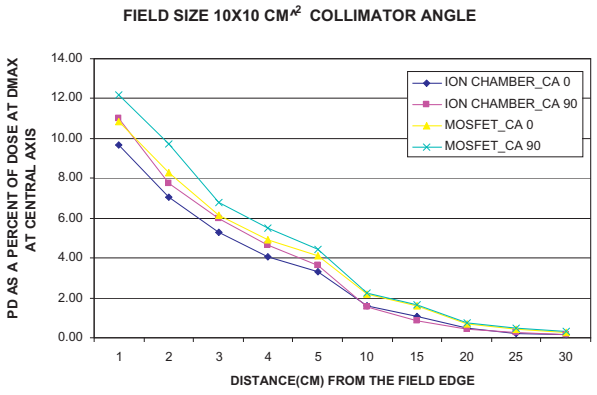

Fig. 4. Peripheral dose (PD) in phantom for 6 MV X-rays measured at 10 cm depth using ionization chamber and MOSFET detector for field size 10 x 10 cm^2 for collimator angle zero and 90°

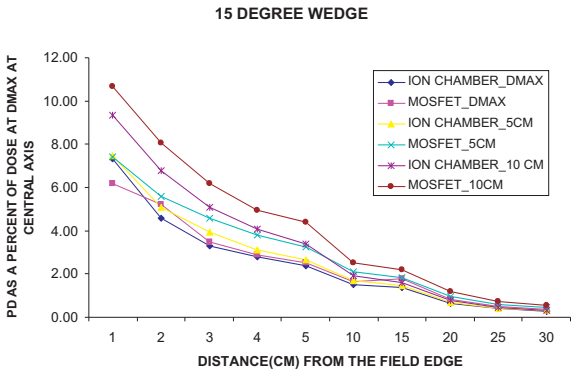

Fig. 5. Peripheral dose (PD) in phantom for 6 MV X-rays measured at dmax, 5 cm and 10 cm depth using ionization chamber and MOSFET detector for 15° wedge with 10 x 10 cm^2 field size

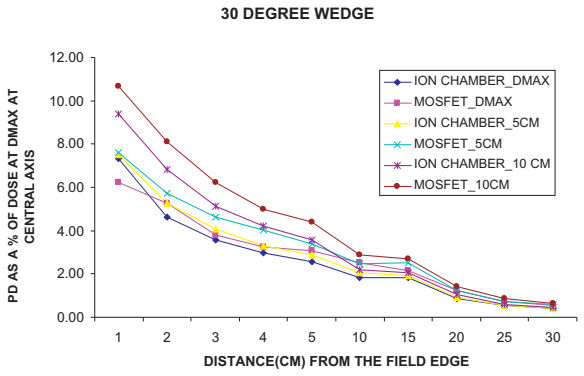

Fig. 6. Peripheral dose (PD) in phantom for 6 MV X-rays measured at dmax, 5 cm and 10 cm depth using ionization chamber and MOSFET detector for 30° wedge with 10 x 10 cm^2 field size

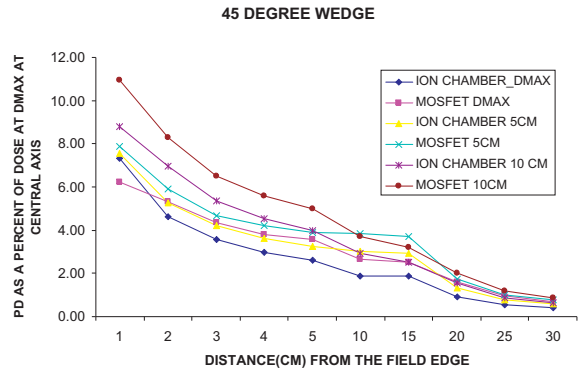

Fig. 7. Peripheral dose (PD) in phantom for 6 MV X-rays measured at dmax, 5 cm and 10 cm depth using ionization chamber and MOSFET detector for 45° wedge with 10 x 10 cm^2 field size

The peripheral dose distributions for 15°, 30° and 45° wedges using the ionization chamber and MOSFET detector are presented in Figures 5, 6 and 7. We delivered 100 MU to measure the peripheral dose for both the open fields and wedge fields. The result also shows a good agreement of dose measured by both detectors for wedged fields.

Peripheral dose distributions for wedged fields were similar in shape to open field dis-

tributions but higher in magnitude. Due to the small wedge factor, the required MU for wedge fields was almost four times larger than for equivalent open fields. Therefore it was expected that wedged PD distributions would be higher in magnitude than for open fields. For distances closer than 30 cm from the field edge, where internal scatter dominates, wedged PD distributions were comparable in magnitude with open field distributions. At larger distances from the field edge, where leakage radiation dominates, wedged field PD distributions become larger in magnitude, approaching the ratio of monitor units required to deliver the same dose with wedge and open fields.

The MOSFET correction for various field sizes and depths with wedges are presented in Table 1. The correction factor was determined as the ratio of the readings between the ionization chamber and MOSFET detector. The variation in the correction factor at different distances from the field edges for the field size and depth was represented as the standard deviation in the brackets. Hence by applying the calibration factor and the appropriate correction factor to the MOSFET readings the peripheral doses are estimated accurately with their respective standard deviations by using MOSFET detectors.

The measured calibration factor for the high sensitivity MOSFET detector in high sensitivity mode was 7.77mV/cGy.

The MOSFET detector can be used to estimate the actual peripheral dose received by the patient during the course of radiotherapy treatment. The dose measured by the MOSFET in the patient is estimated as follows:

PD (cGy) = MOSFET reading (mV) x Calibration factor (cGy/mV) x MOSFET correction factor

The MOSFET correction factor for the various field sizes, beam modifiers and at different depths can be obtained from Table 1. The MOSFET correction factors are derived by taking the average of the correction factors at different points from the field edge. It is advisable for the users to derive their own MOSFET correction factors. The use of an appropriate correction and calibration factor applied to the MOSFET reading measured in the patient will determine the PD, and the accuracy of PD measured by MOSFET is considered to be adequate for risk assessment.

DISCUSSION

Marilyn Stovall et al. [1] presented data and techniques that allow the medical physicist to estimate the radiation dose that the fetus could receive and to reduce this dose with appropriate shielding. Beam data were presented for a variety of photon beams, including cobalt-60 gamma rays and X-rays from 4 to 18 MV. Designs for simple and inexpensive to more complex and expensive types of shielding equipment were described. Clinical examples show that proper shielding can reduce the radiation dose to the fetus by 50%.

Robin L. Stern [2] measured the peripheral dose at two depths and two field sizes for 6 and 18 MV photons from a linac with an MLC. The MLC was configured both with leaves fully retracted and with leaves positioned at the field edges defined by the secondary collimator jaws. Comparative measurements were also made for 6 MV photons from a linac without

Table 1. Correction factor of MOSFET detector with respect to ionization chamber

Field size (cm^2)	Correction factor at dmax (SD)	Correction factor at 5 cm (SD)	Correction factor at 10 cm (SD)
5 x 5	0.83 (8.1)	0.75 (7.6)	0.69(8.9)
10 x 10	0.75 (8.5)	0.71(11.4)	0.74(12.6)
20 x 20	0.76 (9.2)	0.75 (7.8)	0.75(11.1)
10 x 10 CA 90°	0.89 (18.0)	0.77(13.7)	0.74(11.5)
10 x 10 15° wedge	0.86 (8.6)	0.79 (7.4)	0.74(9.4)
10 x 10 30° wedge	0.82 (8.4)	0.81 (6.6)	0.78(5.9)
10 x 10 45° wedge	0.76(6.2)	0.81 (6.2)	0.79(3.5)

an MLC. Peripheral dose was determined as a percentage of the central axis dose for the same energy, field size, and depth using diode detectors in solid phantom material. They found that data for the 6 MV without MLC agreed with those for the beam with MLC leaves retracted. For both energies at all depths and distances from the field edge, configuring the MLC leaves at the field edge yielded a reduction in peripheral dose of 6% to 50% compared to MLC leaves fully retracted.

Eric E Klein et al. [3] conducted slab phantom peripheral dose measurements for very small field sizes (from 2 to 10 cm). They collected the data at distances ranging from 5 to 72 cm away from the field edges. 6 MV, 120-leaf MLC Varian axial beams were used. A phantom mimicking a 3-year-old was configured. Micro (0.125 cc) and cylindrical (0.6 cc) ionization chambers were appropriated for the thyroid, breast, ovaries, and testes. The PD was recorded by electrometers set to the 10^{-10} scale. For the slab phantom studies, close peripheral points were found to have a higher dose for low energy and larger field size and when the MLC was not deployed. For points more distant from the field edge, the PD was higher for high-energy beams. MLC orientation was found to be inconsequential for the small fields tested. Peripheral dose in close proximity (<10 cm) to the field edge was dominated by internal scatter; therefore, field-size differences overwhelmed phantom size effects and increased MU. Distant peripheral dose, dominated by head leakage, was higher than predicted, even when accounting for MU (factor of 3), likely due to the paediatric phantom size. PD to OAR for paediatric IMRT cannot be predicted from large-field full phantom studies. For regional OAR, doses were likely lower than predicted by existing "large field" data, while the distant PD was higher.

Martin J Butson et al. [4] evaluated the accuracy of a MOSFET dosimetry system with respect to peripheral therapeutic doses from high-energy X-rays. The results were compared with ionization chamber measurements in the same peripheral regions of the beam. For 6 MV and 18 MV X-ray beams, the MOSFET system in the high-sensitivity mode produces reproducibility of dose measurement with relative standard deviations within 1% of the maximal dose in the beam, if the measurement was made up to 15 cm from the beam edge. The results showed that the MOSFET device can adequately measure peripheral doses, which would be beneficial for in vivo dose assessments in radiotherapy.

Sasa Mutic et al. [7] performed the measurement to evaluate PD distributions for a linear accelerator equipped with a secondary MLC, backup diaphragms, and universal wedge (UW). Measurements were made with an ionization chamber inserted into a 20 x 40 x 120 cm^3 water-equivalent plastic phantom with the secondary collimator and MLC settings of 5 x 5, 10 x 10, 15 x 15, and 25 x 25 cm^2 with and without UW. Data were acquired along the machine's longitudinal axis for 6, 10, and 18 MV photons. Peripheral dose distributions were measured with the collimator rotated to 0° and 270° for open field measurements and to 0°, 180°, and 270° for wedged fields; this allowed evaluation of peripheral dose distributions as a function of collimator rotation. Wedged fields were normalized to deliver the same dose at the depth of maximum dose on the central axis as open fields. The measured PD distributions were generally comparable to data reported by TG-36. At distances close to the field edge less than 30 or 40 cm, the measured PD distributions were lower when the measurement point was shielded by solid jaws than with MLC and backup diaphragm. At longer distances, this trend reversed for all energies and evaluated field sizes. However, the difference in PD distribution with collimator rotation was not large enough to warrant strategic positioning of the collimator to reduce the dose to critical structures outside the primary radiation field. Because internal scatter dominates close to the field edge, wedged PD distributions were comparable to open field doses at distances closer than 30 cm. However, at distances larger than 30 cm from the field edge, wedged PD distributions were significantly greater than those for open fields due to increased contribution of leakage radiation. Increased leakage radiation was due to the increase in wedged field monitor units, which was related to a small wedge factor of 0.27 to 0.29.

D.S. Sharma et al. [8] found that the increase in the number of monitor units in slid-

ing window intensity-modulated radiotherapy, compared with conventional techniques for the same target dose, leads to an increase in PD. PD from a linear accelerator was measured for 6 MV X-ray using a 0.6 cm^3 ionization chamber inserted at 5 cm depth into a 35 cm x 35 cm x 105 cm plastic water phantom. Measurements were made for field sizes of 6 cm x 6 cm, 10 cm x 10 cm and 14 cm x 14 cm, shaped in both static and dynamic multileaf collimation (DMLC) mode, employing strip fields of fixed width of 0.5 cm, 1.0 cm, 1.5 cm, and 2.0 cm, respectively. The effect of collimator rotation and depth of measurement on peripheral dose was investigated for a 10 cm x 10 cm field. Dynamic fields require 2 to 14 times the number of monitor units than does a static open field for the same dose at the isocentre, depending on strip field width and field size. The peripheral dose resulting from dynamic fields manifests two distinct regions showing a crest and trough within 30 cm from the field edge and a steady exponential fall beyond 30 cm. All dynamic fields were found to deliver a higher PD compared with the corresponding static open fields, being highest for the smallest strip field width and largest field size; also, the percentage increase observed was highest at the largest out-of-field distance. For a 6 cm x 6 cm field, dynamic fields with 0.5 cm and 2 cm strip field width deliver PDs 8 and 2 times higher than that of the static open field. The corresponding factors for the 14 cm x 14 cm field were 15 and 6, respectively. The factors by which PD for DMLC fields increase, relative to jaws-shaped static fields for an out-of-field distance beyond 30 cm, were almost the same as the corresponding increases in the number of MU. Reductions of 20% and 40% in PD were observed when the measurements were done at a depth of 10 cm and 15 cm respectively. They concluded that a knowledge of PD from the DMLC field was necessary in a whole body dose and the likelihood of radiation-induced secondary malignancy.

Benedick A Fraass et al. [9] investigated the peripheral dose (PD) for ^{60}Co, 4, 6, and 10 MV X-ray machines. The measurements were carried out down to dose levels of about 0.1% of the peak dose in the beam, since that dose level may be of clinical importance in some situations. The PD measurements for the various machines are qualitatively similar, which allows the identification of a simple basic data set which can characterize the PD for any particular machine. The PD was separated into two components: in-phantom scatter dose and transmission (leakage) dose. Knowledge of the two components is important clinically when shielding is considered. One of the most important parameters in the measurement of PD is the distance from the radiation field edge to the point of measurement. The PD decreases approximately exponentially with distance from the field edge. Leakage and collimator scatter factor magnitude to be known clearly that contribute to the dose outside the field because these components can be reduced by placing a lead shield over the critical area. Near the beam edge the collimator scatter contributes to the PD and the leakage becomes the main contributor at a greater distance from the field edge.

Our data show that the PD increases as the field size increases and this effect is more pronounced closer to the beam edge and is due to the scatter within the phantom from the treatment beam. We also found that in the region of about 10 to 20 cm from the field edge, the collimator scatter decreases so that the major contribution to the PD is due to scatter within the phantom. At about 30 cm, scatter in the patient and head leakage are approximately equal and beyond that point head leakage dominates. It is also clear from our data that as the depth of the measurement point increases the PD also increases because of increased scatter area within the phantom at larger depths. Only a marginal increase in PD is seen due to collimator rotation to 90 degrees, which may be due to scattered photons from the jaws reaching the detector directly instead of interacting with the MLC jaws. An increase in PD while using beam modifiers was observed. All the MOSFET measured data are comparable with the ionization chamber measured data.

In our study the MOSFET dosimeter is found to be a valuable tool to measure the PD, with the advantage of real time measurement.

CONCLUSIONS

A reduction in PD leads to a reduction in complication rates for all normal tissues outside the treatment field. This is especially relevant

for structures that would otherwise receive a dose at or near their respective tolerance levels. The Thomson & Nielson MOSFET dosimetry system provides an adequate assessment in peripheral regions of high-energy X-ray beams. The MOSFET results were in good agreement with ionization chamber data and hence by applying the calibration factor and the appropriate correction factor to the MOSFET detector readings the peripheral doses in individual patients can be estimated. The MOSFET detector has a compact construction, provides instant readout, is of minimal weight, does not involve a connection to a high voltage terminal and can be used on any surface. The accuracy in the estimation of PD using the MOSFET detector is adequate for risk assessment. Because of the above favourable attributes, the MOSFET is extremely good for the measurement of peripheral dose.

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