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PHYSICS CONTRIBUTION

IN VIVO DOSIMETRY USING A LINEAR MOSFET-ARRAY DOSIMETER TO DETERMINE THE URETHRA DOSE IN ^{125}I PERMANENT PROSTATE IMPLANTS

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Purpose: ***In vivo*** **dosimetry during brachytherapy of the prostate with ^{125}I seeds is challenging because of the high dose gradients and low photon energies involved. We present the results of a study using metal-oxide-semiconductor field-effect transistor (MOSFET) dosimeters to evaluate the dose in the urethra after a permanent prostate implantation procedure.**

Methods and Materials: Phantom measurements were made to validate the measurement technique, determine the measurement accuracy, and define action levels for clinical measurements. Patient measurements were performed with a MOSFET array in the urinary catheter immediately after the implantation procedure. A CT scan was performed, and dose values, calculated by the treatment planning system, were compared to ***in vivo*** **dose values measured with MOSFET dosimeters.**

Results: Corrections for temperature dependence of the MOSFET array response and photon attenuation in the catheter on the ***in vivo*** **dose values are necessary. The overall uncertainty in the measurement procedure, determined in a simulation experiment, is 8.0% (1 SD).** ***In vivo*** **dose values were obtained for 17 patients. In the high-dose region (> 100 Gy), calculated and measured dose values agreed within 1.7% ± 10.7% (1 SD). In the low-dose region outside the prostate (< 100 Gy), larger deviations occurred.**

Conclusions: MOSFET detectors are suitable for ***in vivo*** **dosimetry during ^{125}I brachytherapy of prostate cancer. An action level of ± 16% (2 SD) for detection of errors in the implantation procedure is achievable after validation of the detector system and measurement conditions.**

In vivo **dosimetry, MOSFET, Permanent prostate implant, Dose evaluation, Treatment planning system, Iodine-125.**

INTRODUCTION

Brachytherapy using transperineal interstitial permanent prostate implants (TIPPB) administers a highly conformal dose to the tumor volume. However, uncertainties in seed positioning during the implantation procedure, errors in needle positioning, seed migration, or variations in seed activity may cause deviations from the prescribed dose. Several studies evaluated the dose in the prostate and organs at risk (OARs) after TIPPB, using a postimplant CT scan and the treatment planning system (TPS) to calculate the absorbed dose (1). With this method, errors in dose calculation, such as interseed attenuation, compensation for tissue inhomogeneities, and errors in seed activity, are not easy to detect (2–4). In addition, seeds with a short interseed distance are difficult to recognize on CT images. Inaccuracies in seed registration will result in imprecise dose information. *In vivo* dosimetry may provide useful information about the actual dose to the urethra (5). It is useful to evaluate the implantation technique, including the accuracy of the TPS, which may offer the ability to link dose distributions in OARs to clinical side effects. Nevertheless, *in vivo* dosimetry during brachytherapy is challenging because of the high dose gradients and the low energy of the photons emitted by the seeds.

The aim of this study was to validate the new linear 5-metal-oxide-semiconductor field-effect transistor (MOSFET) array dosimeter for dose verification after permanent ^{125}I prostate implants to evaluate the dose in the urethra.

METHODS AND MATERIALS

^{125}I brachytherapy sources (Intersource^{125}, International Brachytherapy, Seneffe, Belgium) with a mean nominal activity of 0.6 U

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(0.5 mCi) were used. The activity of each seed in a batch was within ± 4% of the mean nominal activity provided by the manufacturer.

Our clinically applied TPS (PSID 4.1, International Brachytherapy), designed for real-time dynamic ultrasound-guided radiation therapy of prostate cancer, was used for dose calculation. This system complies with the American Association of Physicists in Medicine Task Group No. 43 (TG-43) recommendations (6). Patient planning is based on the criteria that the 145-Gy isodose line covers the prostate volume, the urethra dose does not exceed 200 Gy, and the medium rectum volume exceeding 100 Gy should be < 8 cc (7).

For dose measurements, a high-sensitivity MOSFET array (Linear 5ive MOSFET Array, TN-502LA5, Best Medical Canada, Ottawa, Canada) at the high-sensitivity-bias setting was used. The array contains five MOSFET dosimeters, which are 4 mm long, 1.8 mm wide, and 1.3 mm thick, all fixated on a 46-cm flexible cable, with 2-cm inter-MOSFET spacing. The Mobile MOSFET Dose Verification System (TN-RD70W, Best Medical Canada, Ontario, Canada) was used for online dose readout on a remote PC.

Phantom measurements

Angular dependence of the MOSFET-array response. Studies performed with single dosimeters (5, 8) and with the linear array at ^{60}Co energies (9) demonstrated a small angular dependence. To test the angular response of the linear MOSFET array for ^{125}I sources, the detector was placed at the center of a polymethylmethacrylate (PMMA) phantom made in-house (Fig. 1a). Three strengths, each containing of three seeds, were placed 0.5, 1, and 1.5 cm from the detector to obtain higher dose-rates and more accurate measurements. The direction from the MOSFET to the top of the black epoxy bulb of the detector was defined as 0°. The angle of the detector was varied in steps of 30° from 0° to 330°. For each angle, five measurements were obtained, and the results were averaged. The result of each angle was related to the mean result of all angles.

Calibration. The calibration coefficient, which converts the radiation-induced voltage shift of the detector into dose, decreases with increasing energy in the keV range (10). Therefore, the energies used for the clinical measurements should be used to assess the calibration coefficient in this low energy range. To determine the calibration coefficient for each MOSFET of the linear array, 12 ^{125}I seeds were positioned circularly around the detector using a PMMA calibration phantom (8 × 8 × 15 cm) made in house, and the MOSFET array was placed at the center (Fig. 1b). This setup was used to simulate the clinical situation, reduce measurement time, and reduce uncertainties in the position of the detector because the dose distribution in the area of the measurement is almost homogeneous. The dose rate was calculated using the American Association of Physicists in Medicine TG-43 recommendations. We used a reference distance of 1 cm and the point of interest relative to the source longitudinal axis at the reference angle of 90°. For this type of seed, the dose-rate constant is 1.014 cGyh^{-1}U^{-1} (11). A supplementary correction from water to the material of the calibration phantom of 11% (12) was used.

Corrections were made for source decay during the time interval between calibration and the apparent activity, using the known half-life of ^{125}I.

Each MOSFET was placed in the plane of the seeds, five measurements were performed for each MOSFET, and the results were averaged and compared with the calculated dose values.

Absorption of the catheter material. The *in vivo* measurements were performed with the MOSFET array placed in a catheter. The TPS assumes all material surrounding the seeds is water (*i.e.,* the influence of catheter absorption on the dose distribution is not included). To test the difference in absorption between water and the catheter material, we performed a set of phantom measurements. A PMMA phantom made in-house, similar to the calibration phantom but with a larger hole in the center of the phantom, was used. The larger hole allowed for catheter placement. A second ring containing 12 holes for seed placement was positioned at a 2-cm distance from the detector to approach clinical dose rates. A Foley catheter, a silicone two-way catheter with an outside diameter of 6 mm (18 French) and a length of 40 cm (Ref. 170605, Teleflex Medical, Mijdrecht, The Netherlands), was positioned in the center of the phantom. The MOSFET array was placed in the catheter, and 24 strands, each containing three seeds, were placed around the detector. The phantom was placed in a water tank to fill the space between the MOSFET and catheter material with water (Fig. 1c). The position of the array was determined accurately (± 0.5 mm), having MOSFET 2 positioned in the plane of the center of the strands. The MOSFET position was related to the marks on the flexible cable of the array, relative to the top of the phantom, and defined as MOSFET position 1 (MP1).

To investigate the absorption of the catheter material in case the radiation passes the catheter under an oblique angle, the MOSFET array was pulled down along the catheter with a distance of 8.8 mm (MP2), 17.6 mm (MP3), 25.1 mm (MP4), 33.9 mm (MP5), and 42.7 mm (MP6), again by using the marks on the flexible cable in relation to the top of the phantom. For each position, five measurements were performed. The results were related to an equal set of measurements in the same phantom positioned in a water tank, but with water occupying the space where the catheter material was previously. The results are presented as the ratio of the MOSFET reading attenuated by catheter material compared with the attenuation of the same layer of water.

Temperature dependence of the MOSFET-array response. Ramaseshan *et al.* (13) described the temperature response of micro-MOSFETs, demonstrating a 0.5% variation when increasing the temperature from 20 to 40 °C. The temperature dependence of the new MOSFET array was not yet described. To test the variation in MOSFET array sensitivity due to temperature changes, the large-hole phantom (without the catheter material) and 12 single seeds were used. The temperature sensor was placed inside the hole close to the MOSFET array without shielding the detector from the radiation. The phantom was positioned into a water tank, and the water temperature was varied between 19 and 40 °C. MOSFET 3 was positioned in the plane of the seeds. Measurement times of 10 min were used. The temperature was determined at the start and the end of the measurement, and the mean temperature was used for the results. The results were related to initial calibration conditions at room temperature (20 °C).

Total measurement uncertainty. The reproducibility of the MOSFET detector depends on the energy of the radiation and the integrated dose measured by the MOSFET detector. Lavallee *et al.* (10) reported uncertainties of about 6% (1 SD) using an orthovoltage beam with a mean energy of 30 keV and dose values of about 55 cGy. During the clinical measurements, the uncertainty may be even larger because of the lower signal and positional inaccuracies of the detector. To estimate the total uncertainty, a simulation of a complete treatment was performed. A tank, containing a urinary catheter, was filled with gelatin. Needles were inserted through the holes in the wall of the phantom. We planned a needle distribution, mimicking a clinical plan, using 19 needles and 57 seeds. After the placement of the seeds, the MOSFET array was placed in the urinary catheter, and dose values were obtained (position 1). Five