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In Vivo Dosimetry of an Anthropomorphic Phantom Using the RADPOS for Proton Beam Therapy

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Abstract

The radiation positioning system (RADPOS) combines an electromagnetic positioning sensor with metal oxide semiconductor field-effect transistor (MOSFET) dosimetry, enabling simultaneous online measurement of dose and spatial position. Evaluation points can be determined with the RADPOS. The accuracy of *in-vivo* proton dosimetry was evaluated using the RADPOS and an anthropomorphic head and neck phantom. MOSFET doses measured at 3D positions obtained with the RADPOS were compared with treatment plan values calculated using a simplified Monte Carlo (SMC) method. MOSFET responses, which depend strongly on the linear energy transfer of the proton beam, were corrected using the SMC method. The SMC method was used to calculate only dose deposition determined by the experimental depth-dose distribution and lateral displacement of protons due to the multiple scattering effect in materials and incident angle. This method thus enabled rapid calculation of accurate doses in even heterogeneities. *In vivo* dosimetry using the RADPOS, as well as MOSFET doses, agreed with SMC calculations in the range of −3.0% to 8.3%. Most measurement errors occurred because of uncertainties in dose calculations due to the 1-mm position error. The results indicate that uncertainties in measurement position can be controlled successfully within 1 mm when using the RADPOS with *in-vivo* proton dosimetry.

Keywords

RADPOS, MOSFET, *In Vivo* Dosimetry, Position Sensor, Proton Beam Therapy

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1. Introduction

Proton beam therapy (PBT) provides a therapeutic gain for deeply seated tumors because the depth-dose distribution is characterized by a slowly rising dose near the entrance, followed by a sharp increase near the end of the range. To deliver a highly conformal dose to a tumor while sparing surrounding normal tissue, accurate dose delivery is essential. Proton dose distributions must thus be evaluated accurately, namely by *in vivo* proton dosimetry.

In vivo dosimetry is generally performed by placing some type of detector on the point of interest in the patient anatomy. Therefore, *in vivo* dosimetry requires a very small and easily localized detector, and diode [1], plastic scintillation [2] and thermoluminescent dosimeter [3] are used as an *in vivo* dosimeter. Especially, metal oxide semiconductor field-effect transistor (MOSFET) [4] is useful for patient dose measurements. Because the MOSFET is direct reading with a very small active area (0.04 mm^2), and the physical size of the MOSFET is less than 4 mm^2. In addition, the post radiation signal is permanently stored and is dose rate independent. Then, the reading procedure is fast and simple. The MOSFET has been examined thoroughly [5]-[7].

In proton dose measurements, these detectors have Linear Energy Transfer (LET) dependence. Therefore, we must consider very carefully for quantitative proton dose evaluations with these detectors. On the other hand, the use of *in vivo* dosimetry in proton by measuring the decay of radiation-induced radionuclides has been also studied [8]. Here, Kohno *et al*. challenged *in vivo* dosimetry using the MOSFET as a detector, which can detect directly radiation, in an anthropomorphic phantom for PBT [9]. They reported that large measurement errors are unavoidable because accurate measurement of point doses is difficult.

To improve the accuracy of point dose measurement in *in vivo* dosimetry, the radiation positioning system (RADPOS) was developed. The RADPOS consists of a MOSFET dosimeter coupled with an electromagnetic positioning device. It can be used to simultaneously monitor detector position and measure dose [10]-[12].

Kohno *et al*. investigated the application of the RADPOS in PBT and found that interfering materials, such as metal components of the beam-delivery system's snout, distorted the RADPOS' transmitted field [13]. Although they reported that special attention is needed when using the RADPOS as a position sensor in PBT, we think that this system should be useful as an *in-vivo* proton dosimeter. The RADPOS in clinical practice has not yet been applied to PBT. To improve on the outcomes reported by Kohno *et al*. [9], we made a novel attempt to perform *in-vivo* proton dosimetry using an anthropomorphic phantom and the RADPOS.

2. Materials and Methods

2.1. RADOPOS

The RADPOS (Best Medical Canada, Ottawa, ON) consists of a MOSFET radiation dosimeter with an active area of 0.04 mm^2 and a small cylindrical electromagnetic positioning device (8-mm length, 1.3-mm diameter). To avoid radiation attenuation and disturbance, the positioning sensor and dosimeter are separated by 8 mm. Dose measurement is based on the difference in threshold voltages ΔVth before and after irradiation. In this study, TN-252RD MOSFET detectors with 0.25-mm oxide thickness and a high-sensitivity bias voltage setting were used.

The RADPOS probe is connected to a mobile MOSFET reader to record the threshold dosimeter voltage. The probe is also connected to a 3D Guidance preamplifier and 3D Guidance tracker (Ascension Technology Corporation, Burlington, VT). The 3D Guidance DC magnetic field transmitter, which is connected to the tracker, generates a pulsed 3D magnetic field with well-defined characteristics. The sensor's response to this magnetic field is monitored by the position tracker and analyzed to determine the x, y, and z coordinates, as well as the probe's azimuth, elevation, and roll rotation angle.

The MOSFET reader and 3D Guidance tracker are connected directly or wirelessly to a host computer. Special software allows the user to record the ΔVth of the MOSFET and the spatial coordinates of the position sensor manually or automatically at user-defined intervals.

2.2. *In Vivo* Dosimetry

In vivo dosimetry was performed using the therapeutic proton beam line at National Cancer Center Hospital East. The beam line employs a dual-ring double-scattering method for PBT [14]. The thickness of the first scatter and shape of the second scatter were determined by the energy of the proton beams. The maximum diameter of the system's irradiation field was 200 mmφ. The 190-MeV proton beam was tested daily to ensure that the proton range was within 0.5 mm [15].

For accurate comparison, MOSFET detector outputs were converted to dose values. Measurements were performed in a PMMA dose calibration phantom [9] [16]. A calibrated 0.6-cc Farmer ionization chamber (FIC, type 30,013; PTW, Freiburg, Germany) and the MOSFET detector were placed along a line perpendicular to the beam axis. The proton energy in the calibration point was 157 MeV, and linear energy transfer (LET) was 0.5 keV/μm. Protons in this point are in the proximal region of the Bragg curve, and the MOSFET detector response has no LET dependence. To obtain the dose calibration factor for the MOSFET detector, the detector and FIC were exposed five times to 200 cGy. The dose calibration factor was determined from the average output.

The dose for the MOSFET detector was obtained as the product of the MOSFET reading (mV), dose calibration factor, and LET correction factor. As the MOSFET response depends strongly on the LET of the proton beam [9] [16], it was corrected using the highly precise simplified Monte Carlo (SMC) method [17] [18]. This method enables calculation only of dose deposition determined by the experimental depth-dose distribution and lateral displacement of protons due to the multiple scattering effect in materials and the incident angle. When applied to a complex anthropomorphic phantom, the SMC method reproduced the measured dose distribution well, satisfying an accuracy tolerance of 3 mm and 3% in the gamma index analysis [19]. The computation time using the SMC method on graphics processing unit architecture under the computer-unified device architecture platform for the clinical cases is around 1 minute [20]. As a result, the SMC method enabled rapid and accurate dose calculation in even heterogeneities.

To evaluate the usefulness of the RADPOS under more realistic conditions, *in-vivo* proton dosimetry was performed using the MOSFET detector with an anthropomorphic phantom (The Phantom Laboratory, Salem, CA, USA; **Figure 1**). The phantom's head and neck region contains representations of complex inhomogeneous tissues, with bone, soft tissue, and various materials and shapes. The phantom was immobilized with a mold and mask. The transmitter was positioned so that the x, y, and z axes corresponded to the head-foot, left-right, and anterior-posterior axes in sagittal, coronal, and transverse planes, respectively (**Figure 1**). The RADPOS origin point was defined as the point 200 mm from the transmitter's center along the x axis. **Figure 2** shows the target area for which a treatment plan was designed. We assumed that the target was rectangular and solid, allowing straightforward evaluation of *in vivo* dosimetry. The isocenter was located at the center of the planning target volume (PTV) region. An irradiation condition for the PTV was determined using a treatment planning system developed in house. This system calculates the dose using the SMC method, and the correction factor for the MOSFET response to account for LET effects. A calculation grid size of 1.172-mm was used for the CT image

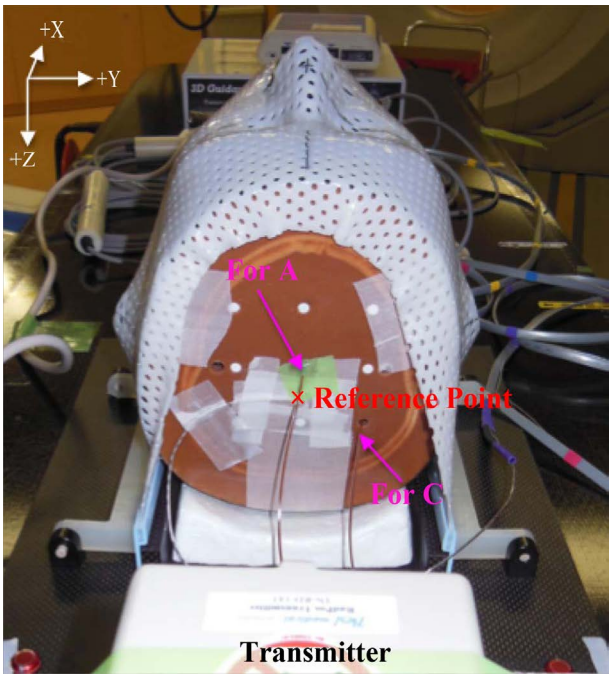

Figure 1. *In-vivo* proton dosimetry using the RADPOS with an anthropomorphic phantom.

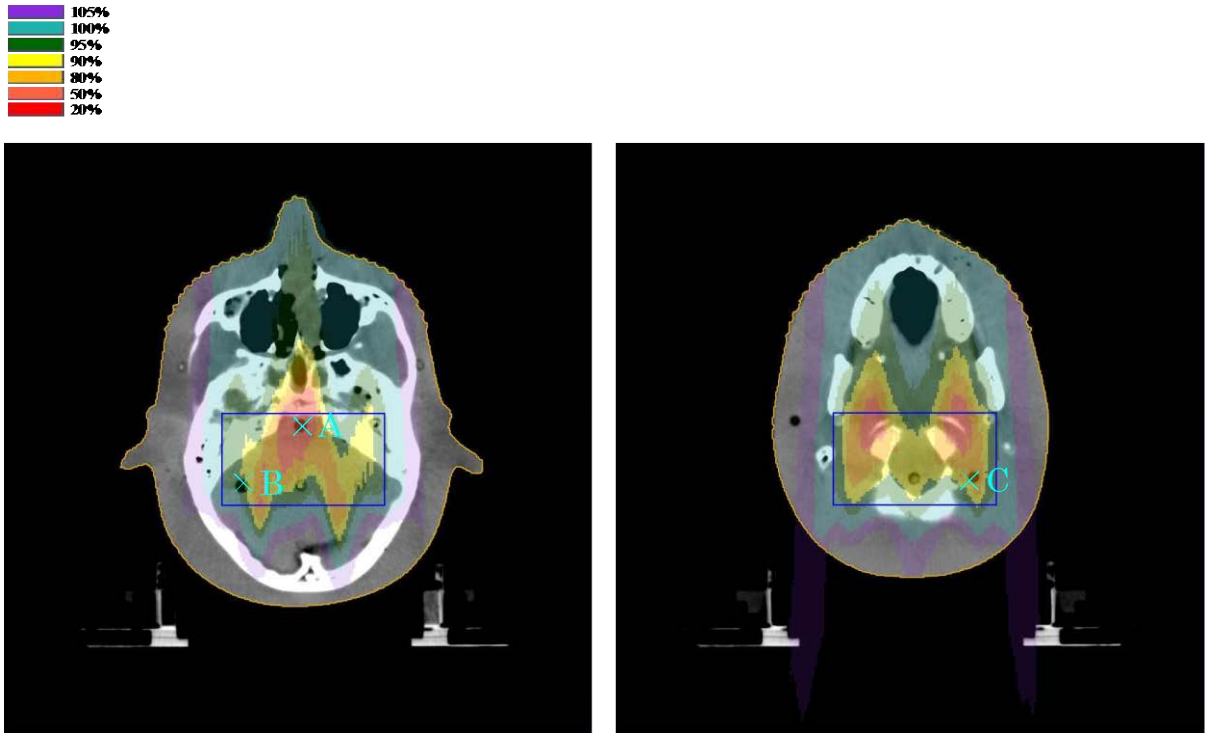

Figure 2. Axial images of the head and neck region in an anthropomorphic phantom, and isodose.

(Asteion; Toshiba Medical Systems, Otawara, Japan). The pixel size of the CT image was based on a 0.586 × 0.586 × 3-mm^3.

The estimated mean statistical error of the calculated dose in the target volume region was within 1% rms. A gantry angle of 0˚ was arranged on the PBT planning system. For the PTV, a bolus and patient collimator were designed using the planning system. A 190-MeV proton beam, a ridge filter of 60-mm spread-out Bragg peak width, and a range shifter of 2.0-mm thickness were selected. As Kohno *et al*. reported that the snout must be located >400 mm from the RADPOS to obtain position accuracy within 1 mm [13], the snout was positioned at −500 mm along the z axis from the isocenter. The isocenter was exposed three times to 200 cGy as a point prescription. Evaluation points are marked on **Figure 2**. Three measurement points were selected to evaluate the dose delivered by protons passing through the inhomogeneities. The dose distribution in the target region was not uniform, but was characterized by steep gradients (up to >5%/mm). Given the presence of complex hot and cold spots around the boundary of inhomogeneity, a precise dose calculation algorithm is desirable in situations involving tissues with significant inhomogeneity.

We measured points A and C (**Figure 2**) three times and point B twice using the RADPOS. To identify each evaluation point, one RADPOS was fixed at the reference point. Using the coordinates of the reference point and each evaluation point, positional relationships were determined. Although *in vivo* dosimetry has conventionally been performed by predetermining evaluation points on a CT image, we obtained these points directly by position measurement with the RADPOS. Thus, we expect that high-precision *in vivo* dosimetry can be performed.

3. Results and Discussion

Figure 3 shows the doses obtained by the uncorrected [MOSFET(−)] and corrected [MOSFET(+)] MOSFET detectors and SMC method at evaluation points A1-C3. The SMC error bar shows estimated maximum and minimum dose errors due to positional uncertainty of ±1 mm. The MOSFET(−) error bar represents the reproducibility of three measurements, and MOSFET(+) includes errors in MOSFET response correction factor calculations with positional uncertainty of ±1 mm.

The SMC results show slight differences in dose (e.g., at points A1-A3) indicating slight differences in the position of the evaluation point. The dose thus appears to change depending on detector position, highlighting the

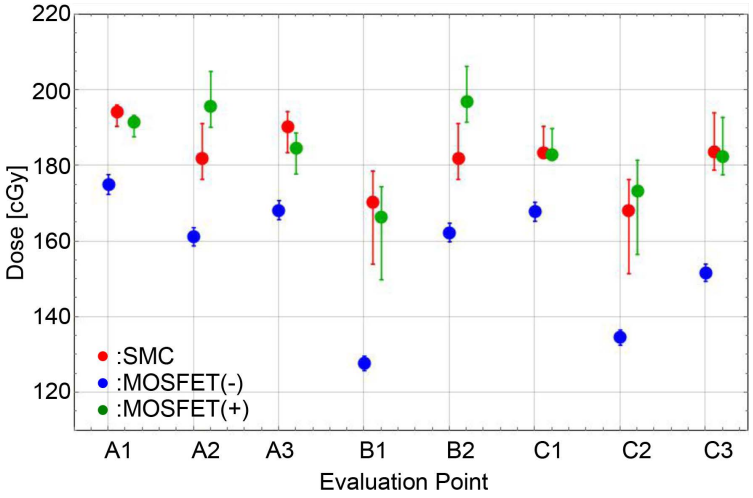

Figure 3. Comparison of doses obtained by the SMC method and uncorrected [MOSFET(−)] and corrected [MOSFET(+)] MOSFET detectors at evaluation points A1-C3.

importance of accurate measurement of this position for *in vivo* dosimetry. MOSFET(−) results deviated significantly from SMC results more than 9%, confirming previous reports of the need for correction of MOSFET(−) data [5] [12]. MOSFET(+) and SMC results deviated less in the range of −3.0% to 8.3%. Most measurement errors occurred because of uncertainties in dose calculation due to the 1-mm position error. Here, in the previous paper [9], Kohno *et al.* estimated considerably large measurement dose errors in a cavity size of 5 mm in diameter due to the MOSFET setup uncertainty in an anthropomorphic phantom. These results mean that the RADPOS could reduce their uncertainty, and play a significant improvement in proton *in-vivo* dosimetery. These findings confirm the usefulness of the RADPOS with a MOSFET detector for *in-vivo* proton dosimetry. However, we deduced slightly large differences of about 4% at points C1 and C2 due to MOSFET angular dependence [16] [21].

4. Conclusion

We evaluated doses delivered in an anthropomorphic phantom using the RADPOS for PBT. The MOSFET doses agreed with SMC calculations within the measurement error. Namely, we could control the uncertainty of the measurement position within 1 mm using the RADPOS with *in-vivo* proton dosimetry. In conclusion, we succeeded in carrying out the precise *in-vivo* dosimetry with the RADPOS. The RADPOS leads a meaningful *in-vivo* proton dosimetry in clinical use. In a future study, we plan to test the clinical application of *in-vivo* proton dosimetry with this RADPOS.

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