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# Experience of using MOSFET detectors for dose verification measurements in an end-to-end $^{192}$Ir brachytherapy quality assurance system

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**ABSTRACT**

**PURPOSE:** Establishment of an end-to-end system for the brachytherapy (BT) dosimetric chain could be valuable in clinical quality assurance. Here, the development of such a system using MOSFET (metal oxide semiconductor field effect transistor) detectors and experience gained during 2 years of use are reported with focus on the performance of the MOSFET detectors.

**METHODS AND MATERIALS:** A bolus phantom was constructed with two implants, mimicking prostate and head & neck treatments, using steel needles and plastic catheters to guide the $^{192}$Ir source and house the MOSFET detectors. The phantom was taken through the BT treatment chain from image acquisition to dose evaluation. During the 2-year evaluation-period, delivered doses were verified a total of 56 times using MOSFET detectors which had been calibrated in an external $^{60}$Co beam. An initial experimental investigation on beam quality differences between $^{192}$Ir and $^{60}$Co is reported.

**RESULTS:** The standard deviation in repeated MOSFET measurements was below 3% in the six measurement points with dose levels above 2 Gy. MOSFET measurements overestimated treatment planning system doses by 2−7%. Distance-dependent experimental beam quality correction factors derived in a phantom of similar size as that used for end-to-end tests applied on a time-resolved measurement improved the agreement.

**CONCLUSIONS:** MOSFET detectors provide values stable over time and function well for use as detectors for end-to-end quality assurance purposes in $^{192}$Ir BT. Beam quality correction factors should address not only distance from source but also phantom dimensions. 

*Keywords:* Brachytherapy; Quality assurance; End-to-end; 192Ir; MOSFET

## Introduction

Experimental quality assurance (QA) of treatment delivery through the measurement of absorbed dose is conducted to a limited extent in brachytherapy (BT). Experimental BT dosimetry is difficult; requirements for positional accuracy are high, and detectors should ideally be of limited size with low dependence on dose, dose rate, and photon energy; see, for example, Baltas et al. (2012), Williamson (2006) (1, 2). Lithium fluoride thermoluminescent dosimetry (TLD) is the hitherto most validated method for use in BT and involved in deriving basic dosimetry data for low energy (<50 keV) sources (3). While solid-state dosimeters like TLDs and electron paramagnetic resonance can be used to measure absorbed doses around $^{192}$Ir BT sources with high accuracy (4−6), these systems require dedicated equipment, extensive workload, and have a delayed signal readout. A recently launched synthetic diamond detector is promising for application to $^{192}$Ir dosimetry as it combines low energy dependence with direct signal readout through use of standard radiotherapy electrometers (7); however, this system is not applicable for *in vivo* purposes.

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Interest in direct reading dosimeters possible to use for *in vivo* dosimetry has raised in recent years (8, 9). Special interest has been on dosimeters capable of online feedback to monitor dose delivery during treatment (10−13). There is one commercially available system, based on a MOSFET (metal oxide semiconductor field effect transistor) detector, available for that purpose (Best Medical [Ottawa, Canada], (14)). This system has been investigated for online *in vivo* dosimetric verification of high-dose-rate $^{192}$Ir BT as inserted into a prostate needle (15) and in the urethra during $^{125}$I seed implantation (16). Problematic for its use in $^{192}$Ir dosimetry are the relatively high atomic number of the active volume ($SiO_2$) and a sensitivity that vary with accumulated dose (14). A linear energy transfer−dependent intrinsic energy dependence that changes with accumulated dose has also been reported (17).

We have used the MOSFET system as part of an end-to-end dosimetric QA system at our clinic for around 2 years. The goal of this work is to report our experiences and so contribute additional knowledge on the system's performance in preparation for possible future *in vivo* applications. The end-to-end system for verification of the $^{192}$Ir BT dosimetric chain is illustrated in Fig. 1 and covers the majority of steps influencing the absorbed doses delivered to patients (computer tomography [CT] acquisition, treatment planning, treatment/measurement); the tests performed in each stage are provided in the bullet lists.

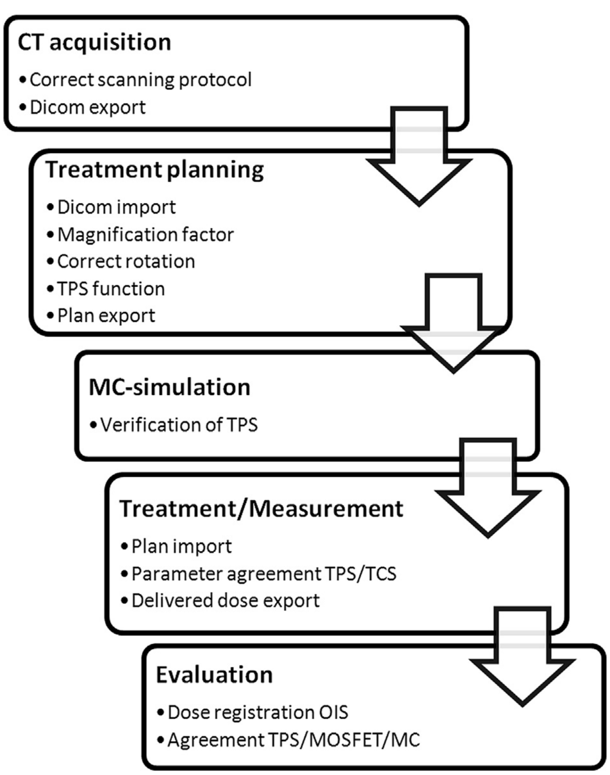


Fig. 1. Graphics describing the separate processes tested by the end-to-end system.

The dose-delivery verification part of the system is currently performed utilizing MOSFETs, and focus of the current work is on the performance of the MOSFET detectors for the task. An evaluation after 2 years of use in conjunction with a total of 56 source exchanges and software upgrades for five afterloaders at our clinic is presented. The calibration of the MOSFET detectors was done in an external beam ($^{60}$Co) as to provide an independent check of the absorbed dose, decoupled from that of the $^{192}$Ir source's air-kerma strength, and the treatment planning system (TPS). A novel experimental investigation regarding the energy dependence of the system is also presented.

## Methods and materials

### *MOSFET detectors and calibration to measure absorbed dose to water*

Three types of MOSFETs are available from Best Medical; singleMOSFET, microMOSFET, and MOSFET array. The 1 mm width of microMOSFET detectors make them suitable for *in vivo* BT measurements as they fit down a 6 F (2 mm diameter) catheter. Single MOSFETs are slightly wider (2.5 mm) and MOSFET array consists of five single MOSFETs placed in a row at intervals of 2 cm. For this work, we have used microMOSFETs and MOSFET array.

The detectors were individually cross-calibrated in a $^{60}$Co external beam against a Farmer chamber with traceability to the Swedish Secondary Standard Dosimetry Laboratory for absorbed dose to water according to the IAEA TRS 398 protocol (18). The calibration phantom was made of PMMA with area of 30 cm × 30 cm, height 15 cm below detectors, and 1 cm above in a 15 cm × 15 cm field. Doses to water in PMMA phantom delivered during calibration were determined by the Farmer chamber and were around 1 Gy. The MOSFETs were positioned with the black epoxy side facing the source.

The lifetime of all MOSFET types is pre-programmed to 20,000 mV. During the 2-year period reported here, a total of seven different microMOSFETs have been used and one MOSFET array (the latter has accumulated approximately 15,000 mV during this period). All the detectors were recalibrated at around 10,000 mV.

### *End-to-end phantom construction*

The phantom is shown in Fig. 2 (schematic, not-to-scale illustration). It was constructed from SuperFlab bolus material (Eckert&Ziegler BEBIG, Berlin, Germany). Two implants were designed, one mimicking a prostate implant using steel needles and the other a head & neck (H&N) implant using plastic catheters. Measurement points for the detectors were prepared both in and outside the implants. For the prostate implant, a plastic catheter large enough to house a MOSFET array was positioned centrally