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Characterisation of a MOSFET-based detector for dose measurement under megavoltage electron beam radiotherapy

W.L. Jong[a], N.M. Ung[a,*], A.H.L. Tiong[b], A.B. Rosenfeld[c], J.H.D. Wong[d,e]

[a] Clinical Oncology Unit, Faculty of Medicine, University of Malaya, 50603 Kuala Lumpur, Malaysia
[b] Nuclear Medicine Centre, Raffles Hospital, Singapore
[c] Centre for Medical Radiation Physics, University of Wollongong, Wollongong, Australia
[d] Department of Biomedical Imaging, Faculty of Medicine, University of Malaya, Kuala Lumpur, Malaysia
[e] University of Malaya Research Imaging Centre, Faculty of Medicine, University of Malaya, Kuala Lumpur, Malaysia

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ABSTRACT

The aim of this study is to investigate the fundamental dosimetric characteristics of the MO*Skin* detector for megavoltage electron beam dosimetry. The reproducibility, linearity, energy dependence, dose rate dependence, depth dose measurement, output factor measurement, and surface dose measurement under megavoltage electron beam were tested. The MO*Skin* detector showed excellent reproducibility (> 98%) and linearity (R^2 = 1.00) up to 2000 cGy for 4–20 MeV electron beams. The MO*Skin* detector also showed minimal dose rate dependence (within ± 3%) and energy dependence (within ± 2%) over the clinical range of electron beams, except for an energy dependence at 4 MeV electron beam. An energy dependence correction factor of 1.075 is needed when the MO*Skin* detector is used for 4 MeV electron beam. The output factors measured by the MO*Skin* detector were within ± 2% compared to those measured with the EBT3 film and CC13 chamber. The measured depth doses using the MO*Skin* detector agreed with those measured using the CC13 chamber, except at the build-up region due to the dose volume averaging effect of the CC13 chamber. For surface dose measurements, MO*Skin* measurements were in agreement within ± 3% to those measured using EBT3 film. Measurements using the MO*Skin* detector were also compared to electron dose calculation algorithms namely the GGPB and eMC algorithms. Both algorithms were in agreement with measurements to within ± 2% and ± 4% for output factor (except for the 4 × 4 cm^2 field size) and surface dose, respectively. With the uncertainties taken into account, the MO*Skin* detector was found to be a suitable detector for dose measurement under megavoltage electron beam. This has been demonstrated in the *in vivo* skin dose measurement on patients during electron boost to the breast tumour bed.

1. Introduction

Compared to megavoltage photon beam, megavoltage electron beam has higher linear energy transfer (LET) and less penetration power (Metcalfe et al., 2007). This results in higher dose to the superficial region and lower dose to the underlying normal tissues, forming a steep dose gradient region. Because of these properties, megavoltage electron beams have always been the choice method to treat skin or near superficial lesions such as radiation boost to the breast tumour bed (Bartelink et al., 2007; Murphy et al., 2011). Typically, electron beams with energies of 4–20 MeV are often used depending on the depth of dose prescription and the dose is deposited within several centimetres from the surface.

Electron beam radiotherapy was administered based on clinical setups using manual calculation, without computerised treatment planning prior to introduction of commercial treatment planning systems (TPS) incorporating electron beam planning capability. The electron dose calculation was originally based on empirical function with assumption of broad beam dose distribution in a homogenous media. Then, pencil beam algorithm based on multiple scattering theory was introduced (Bruinvis and Mathol, 1988; Mah et al., 1989). Recently, a commercial Monte Carlo (MC) dose calculation algorithm for electron beam has become available, which has improved the accuracy of electron dose calculation (Chamberland et al., 2015; Ding et al., 2005; Ojala et al., 2014). Although the electron beam dose calculation algorithms in TPS are available, not all algorithms have comparable accuracies. Further, electron beam radiotherapy without calculation using TPS is still used in clinical practice. In radiotherapy using electron

* Corresponding author.
E-mail address: nm_ung@um.edu.my (N.M. Ung).

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beam, the poor skin sparing effect resulted in higher skin dose. This affects the cosmetic outcome of the electron radiotherapy such as skin reaction and is dependent on the parameters of the electron beam (Murphy et al., 2011) and skin dose (Archambeau et al., 1995).

Based on the reasons stated above, *in vivo* dose measurement is recommended to verify the delivered dose in radiotherapy using electron to prevent errors in delivery. Several dosimetric techniques that are commonly used for *in vivo* dose measurement under electron beam are thermoluminescence dosimeter (TLD) (Anacak et al., 2003; Antolak et al., 1998; Rodriguez-Cortes et al., 2012), optically-stimulated luminescence dosimeter (Nabankema et al., 2016), radiochromic film (Bufacchi et al., 2007; Gamble et al., 2005), silicon diode dosimeter (Marre and Marinello, 2004; Verney and Morgan, 2001; Yaparpalvi et al., 2000) and metal oxide semiconductor field effect transistor (MOSFET) detector (Bloemen-van Gurp et al., 2006; Manigandan et al., 2009).

TLD has high sensitivity and small size, making it a suitable dosimeter for dose measurement in steep dose gradient region, which is encountered in electron beam dosimetry. However, a TLD is not able to provide real-time or instant dosimetry information after radiotherapy. Similarly, radiochromic film does not provide real-time or instant readout although it is able to provide two-dimensional (2D) dose information. In addition, rigorous handling procedures are needed for both TLD and radiochromic film. Semiconductor dosimeters such as a silicon diode have the ability to provide real-time dosimetric information. However, a number of correction factors are needed to account for angular, dose rates and field size dependences (Marre and Marinello, 2004; Yaparpalvi et al., 2000). Moreover, the thick build-up of the silicon diode will perturb the radiation beam and it is more pronounced under electron beam compared to photon beam. Another type of semiconductor dosimeter, the metal oxide semiconductor field effect transistor (MOSFET) detector, has also been used for *in vivo* dosimetry under electron beam (Bloemen-van Gurp et al., 2006; Manigandan et al., 2009). The advantages of the MOSFET detector as an *in vivo* dosimeter include instant readout of the dose, very small sensitive volume, and low dose perturbation.

The MO*Skin* detector, a MOSFET-based detector, was developed by the Centre for Medical Radiation Physics (CMRP) in the University of Wollongong (UoW), Australia. The unique advantage of the MO*Skin* detector over commercial MOSFET is the substitution of epoxy “bubble” encapsulation with a thin film that acts as a build-up for the MO*Skin* detector and gives it a water-equivalent depth (WED) of approximately 0.07 mm in tissue. The MO*Skin* detector has previously been used for dose measurement under photon beam including for skin dose measurement (Hardcastle et al., 2008; Kwan et al., 2008), *in vivo* dose verification during head and neck IMRT (Qi et al., 2009), head and neck serial tomotherapy (Qi et al., 2011), and prostate radiotherapy and brachytherapy (Alnaghy et al., 2015; Carrara et al., 2016; Hardcastle et al., 2010; Kwan et al., 2009; Legge et al., 2017).

In the literature, the MO*Skin* detector has been used and proven to be suitable for *in vivo* dosimetry under photon beam radiotherapy. However, to the best of our knowledge, there has been no application of the MO*Skin* detector for dose measurement under electron beam to date. Therefore, this study investigated the suitability of the MO*Skin* detector for dose measurement under electron beam. The fundamental dosimetric characterisation of the MO*Skin* detector was performed. Specifically, the reproducibility, linearity, energy dependence, dose rate dependence, depth dose measurement, output factor measurement, and surface dose measurement under megavoltage electron beam were tested.

2. Materials and methods

2.1. Experimental setup

All measurements were carried out under a Novalis Tx linear accelerator (Varian Medical Systems, Palo Alto, CA). This linear accelerator has the capability to produce electron beams with nominal energies of 4, 6, 9, 12, 16, and 20 MeV. It is also equipped with applicators that have dimensions of 6 × 6, 10 × 10, 15 × 15, 20 × 20, and 25 × 25 cm^2. The collimator (jaw) setting for the combination of electron beam energy and applicator for this linear accelerator is shown in Table 1.

Table 1
Collimator setting for electron beam. The indicated collimator size is the setting for x and y jaws.

	Collimator size, cm^2					
Applicator, cm^2	**4 MeV**	**6 MeV**	**9 MeV**	**12 MeV**	**16 MeV**	**20 MeV**
6 × 6	20	20	20	11	11	11
10 × 10	20	20	20	14	14	14
15 × 15	20	20	20	17	17	17
20 × 20	25	25	25	25	23	22
25 × 25	30	30	30	30	28	27

All measurements were performed in a motorised water phantom (Blue Phantom[2], IBA Dosimetry, Schwarzenbruck, Germany), except for surface dose measurement which was carried out on the surface of a 30 × 30 × 30 cm^3 solid water phantom (Gammex, Middleton, WI). Unless otherwise stated, all measurements were carried out as per “calibration setup” as shown in Fig. 1, with source-to-surface distance (SSD) of 100 cm, applicator size of 10 × 10 cm^2, and dose rate of 600 MU min^{-1} at depth of maximum dose (6, 13, 21, 28, 34, and 23 mm for 4, 6, 9, 12, 16 and 20 MeV, respectively).

2.2. The MOSkin detector and Gafchromic EBT3 film

The MO*Skin* detector is composed of a hermetically sealed MOSFET sensor with thickness of 350 µm into a Kapton pigtails strip. The MOSFET sensor has gate oxide with volume and thickness of 0.002 mm^3 and 0.55 µm, respectively (Jong et al., 2017). On top of the MOSFET sensor is a thin layer of polyamide film which acts as build-up for the MO*Skin* detector, and gives a 0.07 mm WED. A detailed description of the MO*Skin* detector can be found in Kwan et al. (2008) and Jong et al. (2017).

The MO*Skin* detector was connected to a battery-operated reader

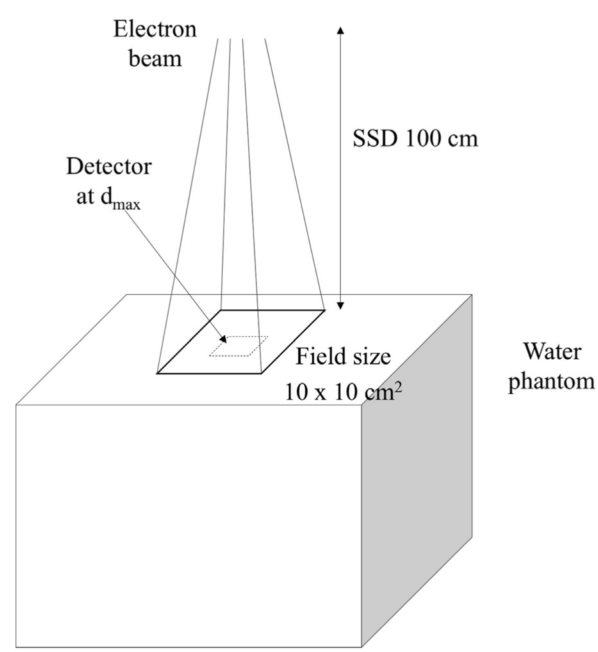

Fig. 1. Calibration setup for the MOSkin detector under megavoltage electron beam.

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