In vivo Dosimetry for Patients Undergoing External Beam Radiation Therapy

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Joanna Cygler – Ottawa Hospital Regional cancer Center
Ben Mijnheer – Netherlands Cancer Center
Slobodan Devic – McGill University Health Center
Indra Das – Indiana University School of Medicine
LEARNING OBJECTIVES

- Usefulness of in-vivo measurements
- Types and characteristics of detectors
- Limitations / Accuracy
Outline of Presentation

• Rationalization – why perform?

• In-Vivo Dosimeter Characteristics
  Real time
  Passive

• Commercially available Dosimeters

• Clinical examples
**CRITICAL REVIEW**

**IN VIVO DOSIMETRY DURING EXTERNAL PHOTON BEAM RADIOThERAPY**

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Table 2. Number of deviations observed between measured and prescribed dose larger than 5%, for the studies mentioned in Table 1

<table>
<thead>
<tr>
<th>Ref.</th>
<th>No. patients</th>
<th>No. deviations</th>
<th>Reason for deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>(38)</td>
<td>1991</td>
<td>14</td>
<td>Erroneous calculation of the dose for irregular fields</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4</td>
<td>Erroneous calculation for isocentric instead of SSD treatment</td>
</tr>
<tr>
<td></td>
<td></td>
<td>11</td>
<td>Drifted output of the treatment unit*</td>
</tr>
<tr>
<td></td>
<td></td>
<td>2</td>
<td>Compensators erroneously placed in tray: gross error*</td>
</tr>
<tr>
<td></td>
<td></td>
<td>2</td>
<td>SSD set-up incorrect: gross error*</td>
</tr>
<tr>
<td></td>
<td></td>
<td>1</td>
<td>Wedge filter forgotten: gross error*</td>
</tr>
<tr>
<td>(39)</td>
<td>792</td>
<td>1</td>
<td>Difference in density of the patient</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6</td>
<td>Errors in calculations, sometimes gross errors</td>
</tr>
<tr>
<td>(40)</td>
<td>7519</td>
<td>3</td>
<td>Data mismatch in prescription*</td>
</tr>
<tr>
<td></td>
<td></td>
<td>3</td>
<td>Incorrect input of data in treatment planning system</td>
</tr>
<tr>
<td></td>
<td></td>
<td>46</td>
<td>Data transcription, miscalculation, neglect of shielding blocks</td>
</tr>
<tr>
<td></td>
<td></td>
<td>7</td>
<td>Wrong or missing shielding blocks or wedges*</td>
</tr>
<tr>
<td></td>
<td></td>
<td>19</td>
<td>Incorrect monitor unit setting*</td>
</tr>
<tr>
<td></td>
<td></td>
<td>1</td>
<td>Mechanical failure of a timer on a cobalt unit*</td>
</tr>
</tbody>
</table>

* Indicates that this error could not have been traced by means of an independent dose calculation program instead of *in vivo* dosimetry. A gross error is an error larger than 10%.
Can we afford not to implement in vivo dosimetry?

In his 2006 report, the Chief Medical Officer, Sir Liam Donaldson, recommends that in vivo dosimetry should be routine [1]. This comes after previous safety initiatives that commenced with “Organization with a memory” [2] and the follow-up report “Building a safer NHS for patients” [3]. He has also initiated change internationally and chairs the World Health Organization (WHO) Patient Safety Initiative [4], which has now established a working party to address some of the clinical risks in the pathway of care for radiotherapy.
Countries requiring in-vivo dose measurement during treatment

- England \(^1\)
- France \(^2\)
- Sweeden \(^3\)

\(^1\) The Royal college of Radiologists 2008
\(^2\) Derreumaux et al. 2008
\(^3\) Nyholm 2008
This memo is provided to alert you to regulatory changes the Agency is in the process of promulgating. Due to the increased use of digital imaging and computed tomography, new requirements are proposed for quality assurance procedures and the qualifications and training of the diagnostic imaging physicist. New regulations are also proposed for notifications of radiation therapy medical events (similar to that for radioactive materials in Section 335.1080) and for electronic brachytherapy.

Reports and Notifications of Medical Events, Section 360.120(i)(3)
This new section will require registrants to report to the Agency any administration of therapeutic radiation to a patient that involves the wrong patient, wrong treatment modality or wrong treatment site, as well as any dose delivered that differs by more than 30% of the weekly prescribed dose or 20% of the total prescribed dose. This section is similar to the long existing requirement for reporting such events from the use of therapeutic radioactive materials, except this new requirement applies to treatments delivered with linear accelerators.
Rationale for in vivo measurements

For patient QA it provides an independent verification of the treatment procedure to identify possible errors in:

- calculation
- patient set-up
- data transfer
Real Time
In Vivo Dosimeter
Characteristics
Philosophy of patient-specific QA at NKI-AVL

For each patient:

- Independent monitor unit check (automatically)
- Consistency check of the basic plan properties by the therapists at the treatment machine
- 2D verification of individual IMRT beams
- 3D verification of the total dose distribution of a VMAT plan, by EPID dosimetry in vivo pre-treatment
Patient dose measurement requirements:

1. EPID transmission measurements

2. Conversion of transmission measurements to dose via EPID calibration

3. An algorithm that accounts for attenuation factors and scatter within the EPID and the patient to reconstruct the patient dose
EPID Calibration for Dose Calculation
Two methods

1. Application of empirical models to convert the measured gray scale image of the EPID into a portal dose image using a calibrated ionization chamber in a phantom.

2. Simulation or modeling of the detector response by either Monte Carlo simulation or by an empirical model.
Conclusions-1

• EPID dosimetry is a fast and accurate tool for 2D verification of IMRT and 3D checks of VMAT delivery, both pre-treatment and *in vivo*.

• A number of (serious) errors have been traced by *in vivo* EPID dosimetry that would not have been discovered with pre-treatment verification.

• In order to facilitate the interpretation of the *in vivo* results, integration of 3D EPID dosimetry and cone-beam CT is under development.
Conclusions-2

EPID dosimetry plays an important role in the total chain of verification procedures that are implemented in our department. It provides a safety net for advanced treatments such as IMRT and VMAT, as well as a full account of the dose delivered. The lack of commercial availability of software tools required for its implementation currently limits its clinical use.
Diode as an in-vivo dosimeter

• Advantages:
  – Higher relative sensitivity
  – Quick response – (1 – 10 μs)
  – Good mechanical stability
  – No external bias needed
  – Small size
  – Smaller energy dependence of mass collision stopping power ratios (between silicon and water compared to air and water)

• Disadvantages:
  – Dependence on temperature, dose rate, energy dependence, angle
  – Require an electrical connection during irradiation
Schematics of inherent buildup geometry – flat geometry

Sun Nuclear

Scanditronix
SSD Calibration

Field Size 10x10 cm²
Depth = 10 cm
Ion chamber

Depth = 5 cm
Detector

SSD = 100 cm

M_{raw} \rightarrow TG 51 \rightarrow Dose at d = 10 cm

PDD \rightarrow Dose (water) at Detector location

CF = \frac{D_0(Q)}{Signal(D_0,Q)} \quad \text{cGy/Signal}
<table>
<thead>
<tr>
<th>Diode</th>
<th>6 MV</th>
<th>20 MV</th>
<th>Co-60</th>
</tr>
</thead>
<tbody>
<tr>
<td>Isorad 1 Gold</td>
<td>0.950</td>
<td>0.957</td>
<td>0.988</td>
</tr>
<tr>
<td>Isorad 2 Gold</td>
<td>0.965</td>
<td>0.973</td>
<td>---</td>
</tr>
<tr>
<td>Isorad Red</td>
<td>0.974</td>
<td>0.994</td>
<td>0.987</td>
</tr>
<tr>
<td>SPD</td>
<td>1.002</td>
<td>0.995</td>
<td>---</td>
</tr>
<tr>
<td>EDP30</td>
<td>0.995</td>
<td>0.998</td>
<td>0.994</td>
</tr>
</tbody>
</table>
Dose Rate Dependence

Dose Rate Dependence (n-type)

0.92 0.94 0.96 0.98 1.00 1.02 1.04 1.06 1.08 1.10

Instantaneous dose rate (cGy/s) x 10^4

S/S(4000)

o - Isorad Gold#1 + - Isorad Red (n-type), ◊ - EDP103G, x - EDP203G, * - Isorad-p Red, ▲ - QED Red (p-type), ▽ - QED Blue

▷ - Isorad-3 Gold, < - Veridose Green

x - QED Red (n-type)

(Saini and Zhu, 2004)

June 24, 2009

AAPM Summer School 2009
Energy Dependence for megavoltage photon

(Saini and Zhu, 2007)

Energy Dependence

Normalized Sensitivity vs. Nominal Accelerating Potential (MV)

(Saini and Zhu, 2007)
Temperature coefficient
≈ 0.3% / °C

(Saini and Zhu, 2002)
Angular Dependence

Scanditronix

Sun Nuclear

(Rikner, Thesis, 1983)

(Saini, Thesis, 2007)

* - Isorad-3 Gold, + - Isorad-3 Red, O - QED Gold, x - QED Red
CMRP MOSFET Dosimetry System

MOSkin detectors, thickness 0.07 mm, see Table 29-III

MOSFET Clinical Dosimetry System:
designed and distributed by CMRP

Courtesy of Anatoly Rosenfeld
Wireless setup

Computer

Treatment Room

Wireless Transceiver

2 meters (6ft)

>10 meters (33ft)

15 meters (50ft)

2 meters (6ft)

5V Supply

Power Outlet

Cygler, MOSFET dosimetry, AAPM Summer School 2009
Threshold voltage shift / $\Delta V_T$

- $\Delta V_T$ is a function of absorbed dose
- That function is linear when the MOSFET operates in the biased mode during the irradiation
- Absorbed dose linearity region increases with the increase of the bias voltage

*Soubra, Cygler, Mackay, Med. Phys. 21(4), 567-572, 1994*

*Cygler, MOSFET dosimetry, AAPM Summer School 2009*
Absorbed dose linearity - dual MOSFET-dual-bias

Energy dependence - TN-502-RD


Cygler, MOSFET dosimetry, AAPM Summer School 2009
Temperature dependence

The DVS is calibrated at 37°C for use at body temperature.

DVS dosimeter is approximately 3.3% more sensitive (higher dose reading for same applied dose) when irradiated at 37°C vs. 23°C

A multiplicative correction factor of 1.033 can be used for room temperature phantom measurements

Courtesy of G. Beyer
Effect of accumulated dose

Correction factors for MOSFETs

- Environmental - temperature (no pressure correction)
- Energy dependence
  - beam energy
  - modality (photons, electrons, particles)
- Accumulated dose
- Dose rate
- Field size
- SSD
- Directional dependence
MOSFET detectors
Advantages vs. disadvantages

**Advantages**
- Very small active volume
- Dual-MOSFET-dual bias system eliminates most correction factors
- Instantaneous readouts (on-line dosimetry)
- Permanent dose storage (Can be read multiple times)
- Waterproof
- Efficient in use

**Disadvantages**
- Finite lifetime (~100 Gy)
- Energy dependence
- Temperature dependence for single-MOSFET-detector
- Sensitivity change with accumulated dose for unbiased MOSFETs

Cygler, MOSFET dosimetry, AAPM Summer School 2009
Passive In Vivo Dosimeters
Luminescent In Vivo Dosimeters
TLDs and OSLDs
TG-191
Recommendations on the clinical use of luminescent detectors
TL and OSL Principles

TL = *thermally* stimulated luminescence

OSL = *optically* stimulated luminescence

Conduction band

Valence band

EXPOSURE

Radiation

Radiation sensor (insulating crystal)
TL and OSL Principles

READOUT

Light emission – **TL**
(e.g., blue, UV)

Thermal stimulation (heating)

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Yukihiara and McKeever
Fig. 3 Decay of DOT signal with time at room temperature

Q(t)/Q(t=1min)

time (min)

0.5 Gy
1.0 Gy
2.0 Gy
4.0 Gy
OSLD and OSLD/Dose as a function of dose normalized to their readings at 0.25 Gy
Optical bleaching with 15 W lamp of OSLDs irradiated to various doses

![Graph](image-url)

- OSLD(t) / OSLD(t=0)
- time (sec)
- Doses: 1.0 Gy, 2.0 Gy, 4.0 Gy, 8.0 Gy
### Energy response of TLD and OSLD detectors normalized to their response at 6 MV

<table>
<thead>
<tr>
<th>Modality</th>
<th>Energy MV</th>
<th>Equivalent Energy MeV</th>
<th>(TLD/D)$^Q_{6\text{MV}}$</th>
<th>(OSLD/D)$^Q_{6\text{MV}}$</th>
</tr>
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<tbody>
<tr>
<td>photons</td>
<td>0.250</td>
<td>0.110</td>
<td>1.28</td>
<td>1.71</td>
</tr>
<tr>
<td></td>
<td>“</td>
<td>1.25</td>
<td>1.02</td>
<td>1.04</td>
</tr>
<tr>
<td></td>
<td>“</td>
<td>6</td>
<td>2.4</td>
<td>1.00</td>
</tr>
<tr>
<td></td>
<td>“</td>
<td>18</td>
<td>5.7</td>
<td>0.99</td>
</tr>
<tr>
<td>electrons</td>
<td>6</td>
<td>0.98</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>“</td>
<td>9</td>
<td>0.99</td>
<td>0.99</td>
</tr>
<tr>
<td></td>
<td>“</td>
<td>12</td>
<td>0.99</td>
<td>0.98</td>
</tr>
<tr>
<td></td>
<td>“</td>
<td>16</td>
<td>0.99</td>
<td>0.98</td>
</tr>
<tr>
<td></td>
<td>“</td>
<td>20</td>
<td>0.99</td>
<td>0.98</td>
</tr>
<tr>
<td>protons</td>
<td>100</td>
<td>1.08</td>
<td>1.05</td>
<td></td>
</tr>
<tr>
<td></td>
<td>“</td>
<td>180</td>
<td>1.06</td>
<td>1.06</td>
</tr>
<tr>
<td></td>
<td>“</td>
<td>250</td>
<td>1.08</td>
<td>1.07</td>
</tr>
</tbody>
</table>

$\sigma \approx \pm 5.5\%$, and $\pm 3.8\%$ within 1 SD for kilovoltage and megavoltage irradiations, respectively
Energy dependence of OSL and TLD dosimeters

- nanodot
- Dot
- TLD (0.15 mm)
- TLD (0.89 mm)
OSLD dose response for 15 consecutive irradiations for 7 different absorbed doses (.25, .5, 1, 2, 4, 8, and 16 Gy) with 6 MV photons.

- 0.25 Gy -> 3.75 Gy
- 0.50 Gy -> 7.5 Gy
- 1.0 Gy -> 15 Gy
- 2.0 Gy -> 30 Gy
- 4.0 Gy -> 60 Gy
- 8.0 Gy -> 120 Gy
- 16.0 Gy -> 240 Gy
Increase in background signal with absorbed dose

- 3.5 Gy (0.25 Gy increments)
- 7.0 Gy (0.5 Gy increments)
- 14 Gy (1.0 Gy increments)
- 28 Gy (2.0 Gy increments)
- 56 Gy (4.0 Gy increments)
- 112 Gy (8.0 Gy increments)
- 224 Gy (16 Gy increments)

Poly. (224 Gy (16 Gy increments))
Poly. (112 Gy (8.0 Gy increments))
Poly. (56 Gy (4.0 Gy increments))
Advantages of TLDs

• Wide useful dose range mrad → 10^3 rad (linearity)

• Dose-rate independence 0 → 10^11 rads/s

• Angular independence

• Reusability

• Readout convenience

• Economy

• Availability of different types and sizes

• Automation compatibility

• Accuracy and precision
Disadvantages of TLDs

• Lack of uniformity
• No immediate read-out
  • Fading
  • Light sensitivity
  • Spurious TL
• Memory of radiation and thermal history
  • Reader instability
  • Loss of reading
Advantages of OSLDs

- Easier read-out procedure
- Re-read the detector
- Easier individual identification
- Optical bleaching easier than thermal annealing to remove radiation effects
- Angular independence
- Accuracy and precision
Disadvantages of OSLDs

- encapsulated in light tight plastic housing
- optical bleaching cannot clear all the radiation effects resulting in an increased background signal
- sensitivity changes with accumulated doses

> 20 Gy
Film Dosimetry

• Radiographic
• Radiochromatic
Radiographic Film

- Base (Cellulose nitrate or Polyester) (typically 200 μm)
- Emulsion (10-20 μm; 2-5 mg/cm³)
  - Gelatin (derivative from bone)
  - grain (size: 0.1~3 μm diameter)
    - AgBr (cubic crystal with lattice distance of 28 nm)
    - AgI
    - KI
    - There are $10^9$-$10^{12}$ grains/cm² in an X-ray film
- Coating
  - Very sensitive which may determine X & Y direction uniformity
Film Processing

- **Developing** [(Metol; methyl-p-aminophenol sulphate or Phenidone; 1phenol 3pyrazolidone)]
  - Converts all Ag⁺ atoms to Ag. The latent image Ag⁺ are developed much more rapidly.

- **Stop Bath**
  - dilute acetic acid stops all reaction and further development

- **Fixer, Hypo** (Sodium Thiosulphate)
  - it dissolves all undeveloped grains.

- **Washing**

- **Drying**
Temperature Dependence of Various Films

- Dupont
- Kodak MRM
- Fuji
- Kodak MR5

Optical Density vs. Developer Temperature (degree F)
Dose Rate Dependence

Optical Density

Exposure, R

Advantage of film dosimetry

- Unrivaled spatial distribution of dose or energy imparted.
- Repeated reading of same film: permanent record
- 2-D distribution with single exposure
- Small detector size
- Wide availability: Kodak, Agfa, Fuji, Dupont, CEA
- Large area dosimetry: Especially for electron beam
- Linearity of dose (over a short dose range, OD can be treated linear with dose for most films)
- Dose rate independence (most clinical dose rates)
Film Dosimetry - Caution

- Strong energy dependence (high sensitivity to low energy photons due to photoelectric interactions in grains)
- Film plane orientation with respect to the beam direction
- Emulsion differences amongst films of different batches, films of the same batch or even in the same film
- Densitometer/Digitizer artifacts
Radiochromic Film

Advantages:
• Self developing
• Energy response
Figure 1: Configuration of GAFCHROMIC® EBT Dosimetry Film
Figure 6: Sensitometric response of GAFCHROMIC® EBT in the visual and red color bands
Merits of radiochromic film for in vivo measurements

- Radiochromic film (EBT model):
  - Tissue equivalence
  - Easy to use and handle
  - Energy in-dependence (30 kVp – 1.25 MeV)
  - Relatively fast readout time (6 hours)
  - 2D dosimeter
<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Diode</th>
<th>MOSFET</th>
<th>TLD</th>
<th>OSLD</th>
<th>Film</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dose</td>
<td>+</td>
<td>+</td>
<td>0</td>
<td>0</td>
<td>+</td>
</tr>
<tr>
<td>Accumulated dose</td>
<td>++</td>
<td>++</td>
<td>+</td>
<td>++</td>
<td>++</td>
</tr>
<tr>
<td>Dose rate</td>
<td>+</td>
<td>+</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Energy</td>
<td>+²</td>
<td>+²</td>
<td>+</td>
<td>+</td>
<td>++</td>
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<tr>
<td>SSD</td>
<td>+</td>
<td>++</td>
<td>0</td>
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<tr>
<td>Field size</td>
<td>++</td>
<td>+</td>
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<td>Linearity</td>
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<td>Reproducibility</td>
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<td>0</td>
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<td>++</td>
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<td>Invasive</td>
<td>+</td>
<td>+</td>
<td>0</td>
<td>0</td>
<td>+</td>
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<tr>
<td>Correction factors</td>
<td>++</td>
<td>++</td>
<td>+</td>
<td>+</td>
<td>+</td>
</tr>
<tr>
<td>Dose uncertainty (Estimated 1SD)^3</td>
<td>± 3%</td>
<td>± 5%</td>
<td>± 2%</td>
<td>± 2%</td>
<td>± 3%</td>
</tr>
<tr>
<td>Main advantages</td>
<td>Reproducibility, Immediate readout</td>
<td>Immediate reading, minor fading</td>
<td>No cables, reused after annealing, few corrections</td>
<td>No cables, readout 10 min post irradiation, reused after optical bleaching</td>
<td>2 D dose, resolution, reread, permanent record, various shapes</td>
</tr>
<tr>
<td></td>
<td>Cumbersome calibration, many corrections, cable</td>
<td>Limited lifetime, high cost</td>
<td>Labor intensive, TLD equipment</td>
<td>Short lifetime, dependence on accumulated dose, OSLD equipment</td>
<td>Light sensitive, processing equipment and maintenance, scanning equipment</td>
</tr>
<tr>
<td></td>
<td>Cost, scanning equipment, strict readout protocol</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
In-Vivo Clinical Applications

- entrance and exit doses
- skin dose
- peripheral dose
- tumor dose
- IMRT
- IGRT
- TBI & TSET
- IORT
EPI D dosimetry using a back-projection algorithm

1) calculate plan
2) measure EPID dose
3) reconstruct dose in many planes for all gantry angles
4) compare plan and reconstructed patient dose

Mihnheer et al.
Rectum IMRT: plan transfer error

fraction 1

fraction 2 (plan sent again)

~20% under-dosage in PTV for 1 fraction

Mihnheer et al.

EPI D vs. Plan
Methods to Obtain the midline dose

1. Arithmetic mean of entrance and exit dose
2. Geometric mean of entrance and exit dose
3. Method of Rizzotti et al.
4. Transmission dose using an EPID
Schematic of the Different Doses involved for Photon *in-vivo* dosimetry
Arithmetic Mean

Assumes linear decrease in dose with depth

\[ D_{\text{midplane}} = \frac{(D_{\text{ent}} + D_{\text{ex}})}{2} \]

Additional corrections to improve calculation energy SSD patient thickness
Geometric mean

Assumes an exponential decrease of dose with depth and a linear attenuation, $\mu(r)$, with field size $r$

$$D_{\text{midline}} = \left( D'_{\text{ent}} \cdot D'_{\text{ex}} \right)^{1/2}$$

where the $D'$ doses are corrected for the differences in distance between the midplane and entrance or exit points
Method of Rizzotti et al.

Applies an empirical relation between the ratio of midline and entrance dose and the ratio of exit and entrance dose measurements corrected for:

- energy
- field size
- SSD
- thickness
Transmission Method

The midplane dose is derived from EPID transmission measurements which are converted to exit dose through a convolution model and applying corrections for:

- differences in divergence
- tissue attenuation
- scatter
Table 1. Ratio of calculated and measured isocenter dose in a $10 \times 10$-cm$^2$ field for a homogeneous phantom with a thickness of 20 cm

<table>
<thead>
<tr>
<th>Method</th>
<th>4 MV</th>
<th>8 MV</th>
<th>18 MV</th>
</tr>
</thead>
<tbody>
<tr>
<td>Arithmetic mean</td>
<td>1.112</td>
<td>1.041</td>
<td>1.014</td>
</tr>
<tr>
<td>Geometric mean</td>
<td>0.948</td>
<td>0.969</td>
<td>0.975</td>
</tr>
<tr>
<td>Rizzotti</td>
<td>1.005</td>
<td>0.995</td>
<td>1.004</td>
</tr>
<tr>
<td>Transmission dose</td>
<td>1.007</td>
<td>0.997</td>
<td>0.998</td>
</tr>
</tbody>
</table>

The estimated uncertainty in the ratios is 1.0\% (1 SD) or better.
Table 2. Ratio of calculated and measured isocenter dose in a 10 × 10-cm² field for inhomogeneous phantom A, which simulates the pelvic region

<table>
<thead>
<tr>
<th>Method</th>
<th>4 MV</th>
<th>8 MV</th>
<th>18 MV</th>
</tr>
</thead>
<tbody>
<tr>
<td>Arithmetic mean</td>
<td>1.166</td>
<td>1.083</td>
<td>1.122</td>
</tr>
<tr>
<td>Geometric mean</td>
<td>0.940</td>
<td>0.980</td>
<td>0.978</td>
</tr>
<tr>
<td>Rizzotti</td>
<td>0.999</td>
<td>0.998</td>
<td>1.002</td>
</tr>
<tr>
<td>Transmission dose</td>
<td>1.024</td>
<td>1.012</td>
<td>1.011</td>
</tr>
</tbody>
</table>

The estimated uncertainty in the ratios is 1.0% (1 SD) or better.
Where the Skin Dose should be reported?

- ICRP 23 =>
- Epidermis:
  - Eyelid – 0.05 mm
  - Sole of foot – 1.5 mm

- ICRU 39 and
- ICRP 60 =>

70 µm
What is the Surface Dose?
### Surface Dose = Skin Dose?

<table>
<thead>
<tr>
<th>Dosimeter</th>
<th>Depth scaled by density (mm)</th>
<th>Correction</th>
</tr>
</thead>
<tbody>
<tr>
<td>Attix chamber</td>
<td>0.048</td>
<td>1.062</td>
</tr>
<tr>
<td>Extrapolation chamber</td>
<td>0.069</td>
<td>1.000</td>
</tr>
<tr>
<td>HS GAFCHROMIC® film</td>
<td>0.153</td>
<td>0.850</td>
</tr>
<tr>
<td>EBT GAFCHROMIC® film</td>
<td>0.153</td>
<td>0.854</td>
</tr>
<tr>
<td>XR-T GAFCHROMIC® film</td>
<td>0.157</td>
<td>0.837</td>
</tr>
<tr>
<td>TLD-100 (0.15 mm)</td>
<td>0.185</td>
<td>0.810</td>
</tr>
<tr>
<td>TLD-100 (0.4 mm)</td>
<td>0.496</td>
<td>0.586</td>
</tr>
</tbody>
</table>
Fig. 2. In vivo TLD extrapolation measurements at the entry point of the medial tangential field on a patient undergoing breast treatment with 6 MV photons. The prescribed incident dose at depth of maximum dose was 129 cGy. Entrance and exit dose are shown as well as a comparison between the sum of these measurements and results from detectors left on throughout the treatment. For comparison, the depths of the basal cell layer and the skin vascularisation are also shown.
Quach et al.: Measurement of radiotherapy x-ray skin dose
Med. Phys. 27 .7., July 2000

Fig. 1. The hemicylindrical chest wall phantom used to simulate a tangential breast treatment. A 20×20 cm² x-ray beam with 2.5 cm overshoot was set incident at 100 cm SSD, for some of the radiochromic film readings 1 cm of bolus was added.

Fig. 2. Percentage near surface dose obtained using various detectors including radiochromic (Gafchromic MD 55-2) film, TLDs (extra thin, thin and normal), and a MOSFET detector. Normalized to incident dose at $d_{\text{max}}$ for each detector.
Fig. 2. Percentage near surface dose obtained using various detectors including radiochromic (Gafchromic MD 55-2) film, TLDs (extra thin, thin and normal), and a MOSFET detector. Normalized to incident dose at $d_{\text{max}}$ for each detector.
Conclusions

- Skin Dose is not the same as the Surface Dose!
- Radiation oncologists should specify skin depth when they ask for the skin dose estimate
- For accurate skin dose estimates a proper correction should be applied with any dosimeter
PHYSICS CONTRIBUTION

IN VIVO MEASUREMENTS WITH MOSFET DETECTORS IN OROPHARYNX AND NASOPHARYNX INTENSITY-MODULATED RADIATION THERAPY

SERGE MARCIÉ, PH.D., ELISABETH CHARPIOT, M.SC., RENÉ-JEAN BENSADOUN, M.D., GASTON CIAIS, M.D., JOEL HÉRAULT, PH.D., ANDRÉ COSTA, PH.D., AND JEAN-PIERRE GÉRARD, M.D.

Department of Radiation Oncology, Centre Antoine-Lacassagne, Nice, France

Purpose: To evaluate the feasibility of in vivo measurements with metal oxide semiconductor field effect transistor (MOSFET) dosimeters for oropharynx and nasopharynx intensity-modulated radiation therapy (IMRT).

Methods and Materials: During a 1-year period, in vivo measurements of the dose delivered to one or two points of the oral cavity by IMRT were obtained with MOSFET dosimeters. Measurements were obtained during each session of 48 treatment plans for 21 patients, all of whom were fitted with a custom-made mouth plate. Calculated and measured values were compared.

Results: A total of 344 and 452 measurements were performed for the right and left sides, respectively, of the oral cavity. Seventy percent of the discrepancies between calculated and measured values were within ±5%. Uncertainties were due to interfraction patient positions, intrafraction patient movements, and interfraction MOSFET positions. Nevertheless, the discrepancies between the measured and calculated means were within ±5% for 92% and 95% of the right and left sides, respectively. Comparison of these discrepancies and the discrepancies between calculated values and measurements made on a phantom revealed that all differences were within ±5%.

Conclusion: Our experience demonstrates the feasibility of in vivo measurements with MOSFET dosimeters for oropharynx and nasopharynx IMRT. © 2005 Elsevier Inc.
Fig. 1. The molded mouth plate with the two catheters.

Fig. 2. Image of seeds on a computed tomography slice.
Agreement of measurements with calculations within 5% despite uncertainties due to interfraction patient positions, intrafraction patient movements, and interfractioN MOSFET positions.
In vivo prostate IMRT dosimetry with MOSFET detectors using brass buildup caps

Raj Varadhan,1 John Miller,1 Brenden Garrity,1 and Michael Weber2

Minneapolis Radiation Oncology,1 550 Osborne Road, Fridley, Minnesota 55432; Radiation Therapy Department,2 Methodist Hospital, 6500 Excelsior Blvd., St. Louis Park, Minnesota 55426 U.S.A.

The feasibility of using dual bias metal oxide semiconductor field effect transistor (MOSFET) detectors with the new hemispherical brass buildup cap for in vivo dose measurements in prostate intensity-modulated radiotherapy (IMRT) treatments was investigated and achieved. In this work, MOSFET detectors with brass buildup caps placed on the patient’s skin surface on the central axis of the individual IMRT beams are used to determine the maximum entrance dose \((D_{max})\) from the prostate IMRT fields. A general formalism with various correction factors taken into account to predict \(D_{max}\) entrance dose for the IMRT fields with MOSFETs was developed and compared against predicted dose from the treatment-planning system (TPS). We achieved an overall accuracy of better than \(\pm 5\%\) on all measured fields for both 6-MV and 10-MV beams when compared to predicted doses from the Philips Pinnacle3 and CMS XiO TPSs, respectively. We also estimate the total uncertainty in estimation of MOSFET dose in the high-sensitivity mode for IMRT therapy to be 4.6%.
Table 2. *In vivo* measurements using MOSFETs compared to corrected target doses from the TPS (CMS XiO). All plans delivered with beam energy 10 MV. (CR = 1.038)

<table>
<thead>
<tr>
<th>Patient</th>
<th>Field #</th>
<th>$D_{\text{max}}$ TPS (cGy)</th>
<th>IMRT CF</th>
<th>Total CF</th>
<th>Expected MOSFET dose (cGy)</th>
<th>Measured MOSFET dose (cGy)</th>
<th>% Diff</th>
</tr>
</thead>
<tbody>
<tr>
<td>#1</td>
<td>2</td>
<td>74.6</td>
<td>0.996</td>
<td>1.036</td>
<td>77.3</td>
<td>78.5</td>
<td>1.6%</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>49.0</td>
<td>0.992</td>
<td>1.049</td>
<td>51.4</td>
<td>52.6</td>
<td>2.2%</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>26.7</td>
<td>0.975</td>
<td>1.020</td>
<td>27.3</td>
<td>28.7</td>
<td>5.1%</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>43.7</td>
<td>0.977</td>
<td>1.024</td>
<td>44.7</td>
<td>46.1</td>
<td>3.1%</td>
</tr>
<tr>
<td></td>
<td>6</td>
<td>84.2</td>
<td>1.012</td>
<td>1.053</td>
<td>88.6</td>
<td>91.1</td>
<td>2.8%</td>
</tr>
<tr>
<td>Total</td>
<td></td>
<td><strong>278.2</strong></td>
<td></td>
<td></td>
<td><strong>289.3</strong></td>
<td><strong>296.9</strong></td>
<td><strong>2.6%</strong></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Patient</th>
<th>Field #</th>
<th>$D_{\text{max}}$ TPS (cGy)</th>
<th>IMRT CF</th>
<th>Total CF</th>
<th>Expected MOSFET dose (cGy)</th>
<th>Measured MOSFET dose (cGy)</th>
<th>% Diff</th>
</tr>
</thead>
<tbody>
<tr>
<td>#2</td>
<td>2</td>
<td>61.1</td>
<td>1.018</td>
<td>1.071</td>
<td>65.4</td>
<td>65.2</td>
<td>-0.4%</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>44.7</td>
<td>0.996</td>
<td>1.009</td>
<td>45.1</td>
<td>44.0</td>
<td>-2.4%</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>34.3</td>
<td>1.026</td>
<td>1.044</td>
<td>35.8</td>
<td>35.8</td>
<td>-0.1%</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>39.1</td>
<td>1.004</td>
<td>1.017</td>
<td>39.8</td>
<td>39.3</td>
<td>-1.3%</td>
</tr>
<tr>
<td></td>
<td>6</td>
<td>62.8</td>
<td>0.991</td>
<td>1.069</td>
<td>67.1</td>
<td>70.0</td>
<td>4.3%</td>
</tr>
<tr>
<td>Total</td>
<td></td>
<td><strong>242.0</strong></td>
<td></td>
<td></td>
<td><strong>253.3</strong></td>
<td><strong>254.3</strong></td>
<td><strong>0.4%</strong></td>
</tr>
</tbody>
</table>
In vivo dose verification of IMRT treated head and neck cancer patients
PER E. ENGSTROM, PIA HARALDSSON, TORSTEN LANDBERG,
HANNE SAND HANSEN, SVEND AAGE ENCELHOLM & HA°
KAN NYSTROM°M
Department of Radiation Oncology, The Finsen Center, Copenhagen University Hospital, Copenhagen, Denmark
Abstract
An independent in vivo dose verification procedure for IMRT treatments of head and neck cancers was developed. Results of 177 intracavitary TLD measurements from 10 patients are presented. The study includes data from 10 patients with cancer of the rhinopharynx or the thyroid treated with dynamic IMRT. Dose verification was performed by insertion of a flexible naso-oesophageal tube containing TLD rods and markers for EPID and simulator image detection. Part of the study focussed on investigating the accuracy of the TPS calculations in the presence of inhomogeneities. Phantom measurements and Monte Carlo simulations were performed for a number of geometries involving lateral electronic disequilibrium and steep density shifts. The in vivo TLD measurements correlated well with the predictions of the treatment planning system with a measured/calculated dose ratio of 1.0029/0.051 (1 SD, N/177). The measurements were easily performed and well tolerated by the patients. We conclude that in vivo intracavitary dosimetry with TLD is suitable and accurate for dose determination in intensity-modulated beams.

Acta Oncologica, 2005; 44: 572/578
Figure 1. Photo of the tube containing ten encapsulated TLD rods. The tube extends several centimetres beyond the tungsten “end markers” only for the purpose of the patients’ comfort. The total length of the tube is approximately 30 cm but depends on anatomy and diagnose. Inserted picture (lower left) shows an EPID image with TLD rods (white crosses) interspaced with lead markers. The black arrow marks the isocentre.

P. Engstrom et al., ACTA Oncologica, 44 (2005) 572
Figure 3. TLD readings of two measurements performed on a rhinopharynx patient with one-week interval. The solid lines represent upper and lower tolerance levels (dosimetric + geometric).

P. Engstrom et al., ACTA Oncologica, 44 (2005) 572
Table I. Comparison between OSLD and TLD dose measurements for TBI patients

<table>
<thead>
<tr>
<th>Location</th>
<th>Patient #1</th>
<th>Patient #2</th>
<th>D(cal)/D(meas) Pat 1 : Pat 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>OSLD/TLD</td>
<td>OSLD/TLD</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Head</td>
<td>0.99</td>
<td>0.99</td>
<td>1.04 : 0.99</td>
</tr>
<tr>
<td>Umbilicus</td>
<td>0.97</td>
<td>0.97</td>
<td>0.99 : 1.02</td>
</tr>
<tr>
<td>Feet</td>
<td>0.98</td>
<td>0.97</td>
<td>1.03 : 1.05</td>
</tr>
</tbody>
</table>

\[ \langle \text{OSLD/TLD} \rangle = 0.98 \pm 0.01 \]
Determination of 2 Fields to 12 Fields factor for TSEI using TLDs and OSLDs

Method: 3 TLDs and 3 OSLDs located at 60° intervals on the surface of a 30 cm diameter tissue equivalent phantom

Irradiation conditions: SSD = 410 cm
- 4 mm lexan sheet proximal to phantom
- E = 6 MeV nominal linac energy
- \( E_{\text{eff}} = 3.4 \text{ MeV} \) at phantom surface

Deliver 290 MUs for 12 fields \( \rightarrow 270 \pm 17^0 \) for 6 phantom orientations of
- \( 0^0 – 60^0 – 120^0 – 180^0 – 240^0 - 300^0 \)

<table>
<thead>
<tr>
<th></th>
<th>2 fld Gy</th>
<th>12 fld Gy</th>
<th>12 fld / 2 fld</th>
</tr>
</thead>
<tbody>
<tr>
<td>OSLD</td>
<td>0.360</td>
<td>1.09</td>
<td>3.03 ± 0.11</td>
</tr>
<tr>
<td>TLD</td>
<td>0.378</td>
<td>1.14</td>
<td>3.02 ± 0.11</td>
</tr>
</tbody>
</table>
Table II. Comparison between OSLD and TLD dose measurements for TSEI patients. Values in parentheses are $D_{(cal)} / D_{(measured)}$.

<table>
<thead>
<tr>
<th>Location</th>
<th>Patient #1</th>
<th>Patient #2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>OSLD / TLD</td>
<td>OSLD / TLD</td>
</tr>
<tr>
<td>Forehead</td>
<td>1.02 (1.13)</td>
<td>1.01 (1.18)</td>
</tr>
<tr>
<td>Chest</td>
<td>0.97 (1.10)</td>
<td>0.94 (1.12)</td>
</tr>
<tr>
<td>Anterior</td>
<td>0.97 (1.11)</td>
<td>1.01 (1.06)</td>
</tr>
<tr>
<td>Posterior</td>
<td>1.04 (1.09)</td>
<td>1.03 (1.09)</td>
</tr>
<tr>
<td>Right lateral</td>
<td>1.02 (1.01)</td>
<td>0.97 (0.96)</td>
</tr>
<tr>
<td>Left lateral</td>
<td>1.02 (1.05)</td>
<td>0.97 (0.92)</td>
</tr>
<tr>
<td>Mid thigh</td>
<td>0.97 (1.11)</td>
<td>1.02 (1.04)</td>
</tr>
<tr>
<td>Mid shin</td>
<td>0.97 (1.03)</td>
<td>0.98 (1.05)</td>
</tr>
</tbody>
</table>

$\langle \text{OSLD/TLD} \rangle = 0.99 \pm 0.03$
Out-of-Field Dose

Requirements:

• Sensitivity
• Energy Dependence
• Size

TLDs or OSLDs
CONCLUSIONS

In vivo dosimetry is an effective tool for quality assurance in EBRT

• Verification of patient dose
• Reveal systematic error for individual patient due to:
  patient contour
  patient mobility
  inhomogeneities
  data transfer
• Some treatment procedures (TSET, TBI) require measurements to evaluate patient dose
• Potential regulatory compliance issues
Thank you for your attention

Questions ????
Srivastava and Das *Med Phys* 33:2089, 2006
IMRT in vivo dosimetry


- In some cases, BB’s placed on top of patient’s mask for original CT