In vivo Dosimetry for Patients Undergoing External Beam Radiation Therapy

Chester Reft

University of Chicago Medical Center

Acknowledgements

Timothy Zhu - University of Pennsylvania
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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I. J. Radiation Oncology ● Biology ● Physics Volume 43, Number 2, 1999

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

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

* 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%.

The British Journal of Radiology, 81 (2008), 681–684

COMMENTARY

Can we afford not to implement in vivo dosimetry?

M V WILLIAMS, MD, FRCP, FRCR and 2A McKENZIE, DSc, FBIR

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

To: Diagnostic Imaging Specialists &
Therapeutic Radiological Physicists

From: Donald Agnew, Manager
Electronic Products Section
IEMA

Date: July 5, 2011

Subject: Proposed Changes to Regulations, Parts 320, 360, 410

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

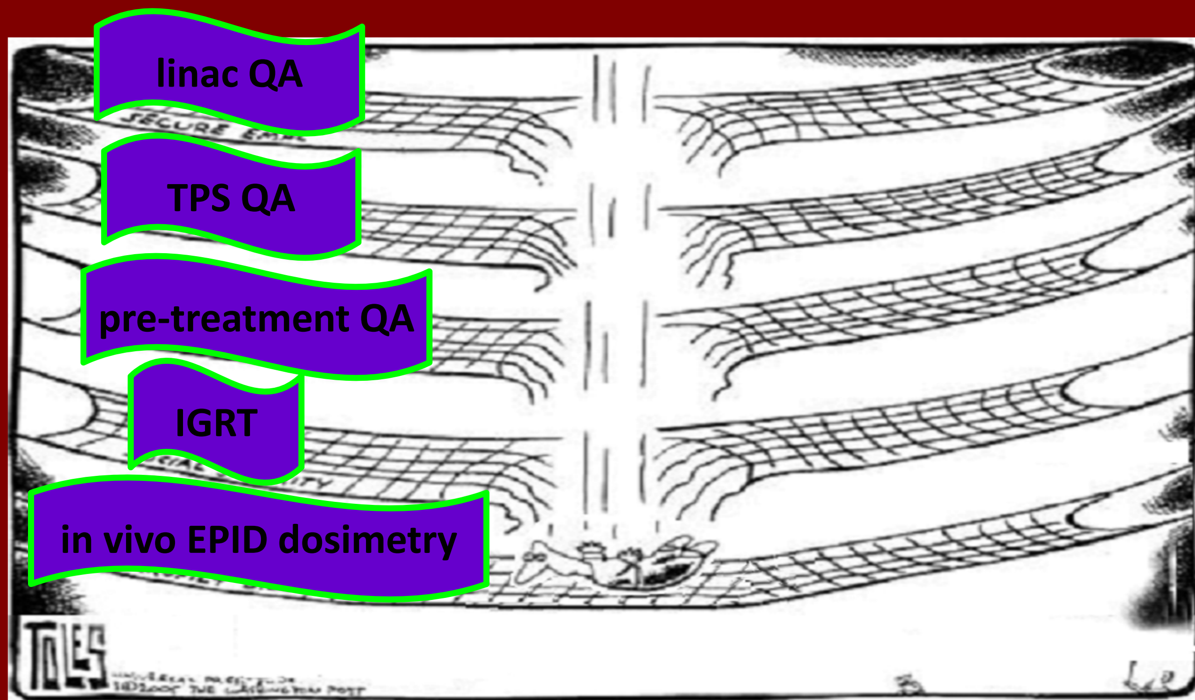
- **EPID dosimetry is a fast and accurate tool for 2D verification of IMRT and 3D checks of VMAT delivery, both pre-treatment and *in vivo*.**

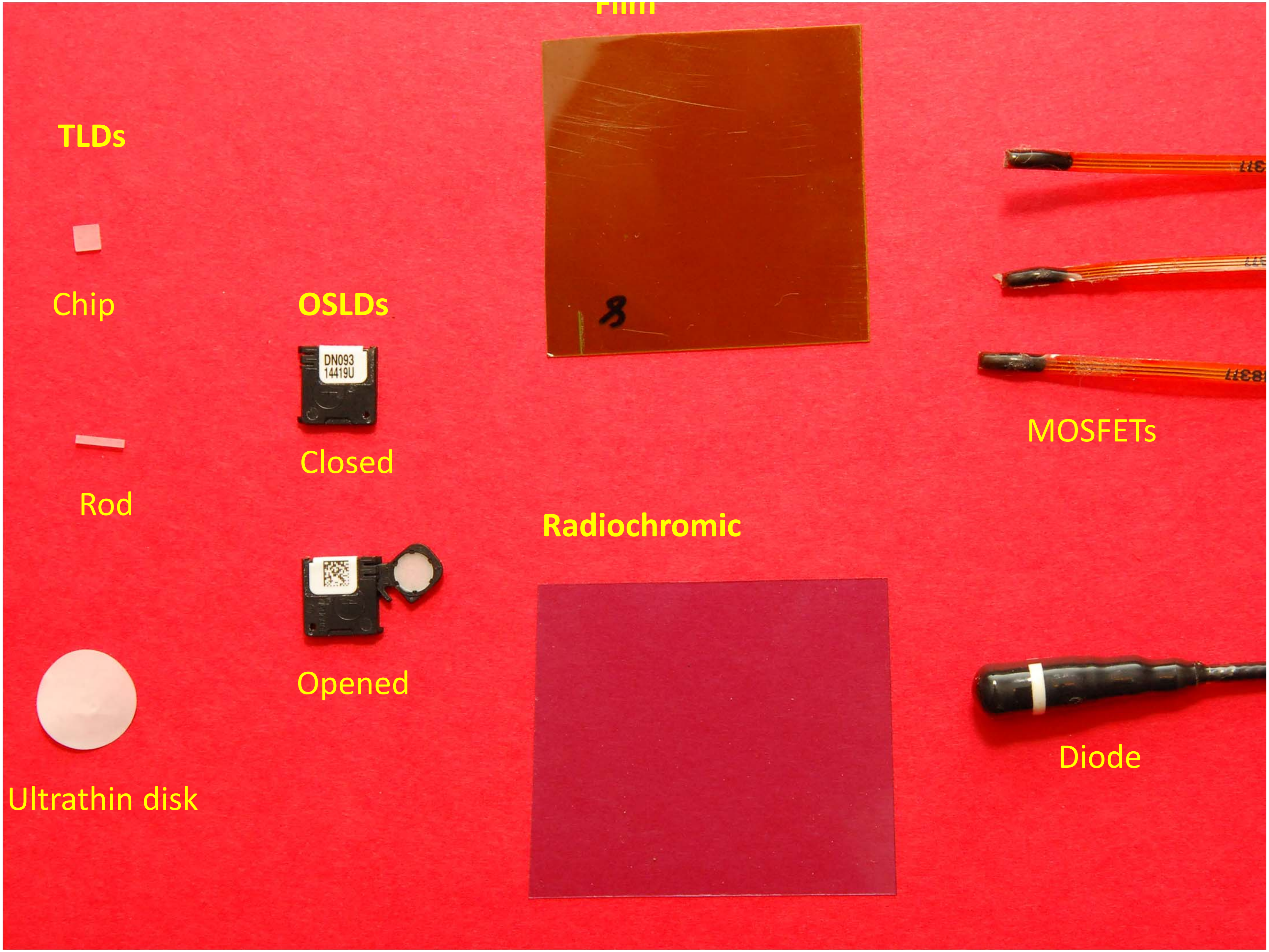
- **A number of (serious) errors have been traced by *in vivo* EPID dosimetry that would not have been discovered with pre-treatment verification.**

- **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.




Film
TLDs
Chip
Rod
Ultrathin disk
OSLDs
DN093
14419U
Closed
Opened
Radiochromic
MOSFETs
Diode

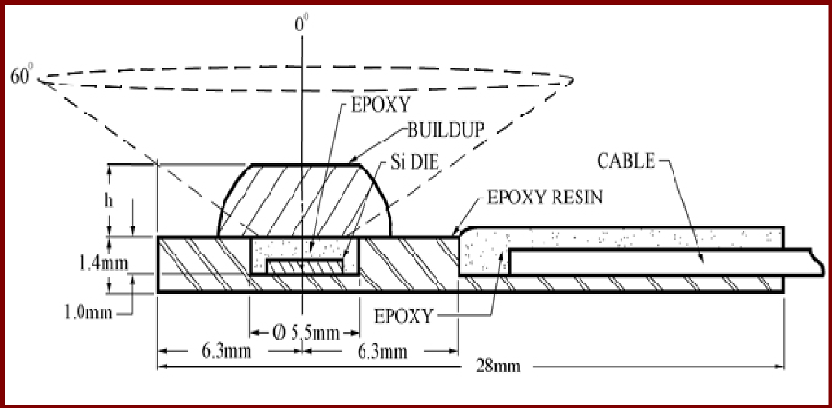
Diode as an in-vivo dosimeter

- **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)

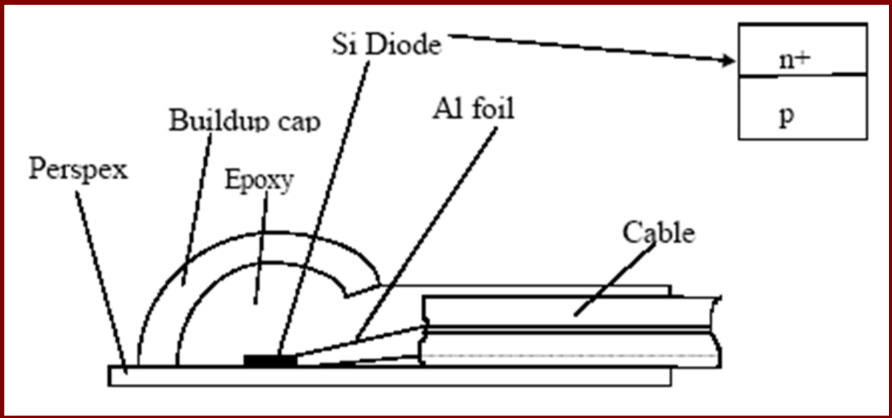
- **Disadvantages:**
  - Dependence on temperature, dose rate, energy dependence, angle
  - Require an electrical connection during irradiation

Schematics of inherent buildup geometry – flat geometry

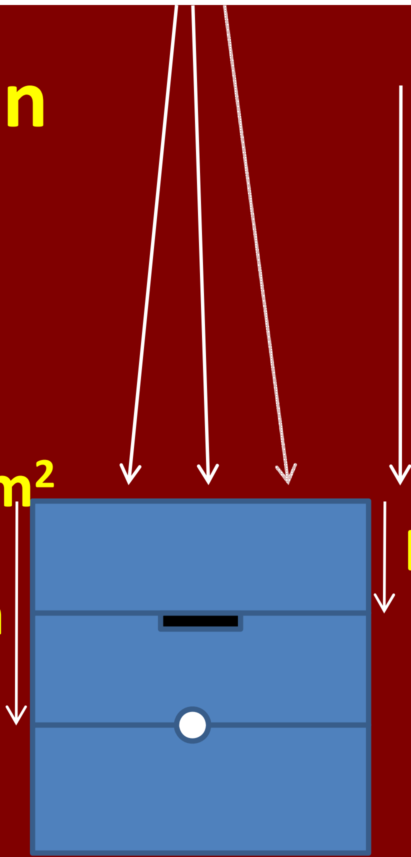
Sun Nuclear



Scanditronix



SSD Calibration



M_{raw} $\xrightarrow{\text{TG 51}}$ Dose at d=10 cm $\xrightarrow{\text{PDD}}$ Dose(water) at Detector location

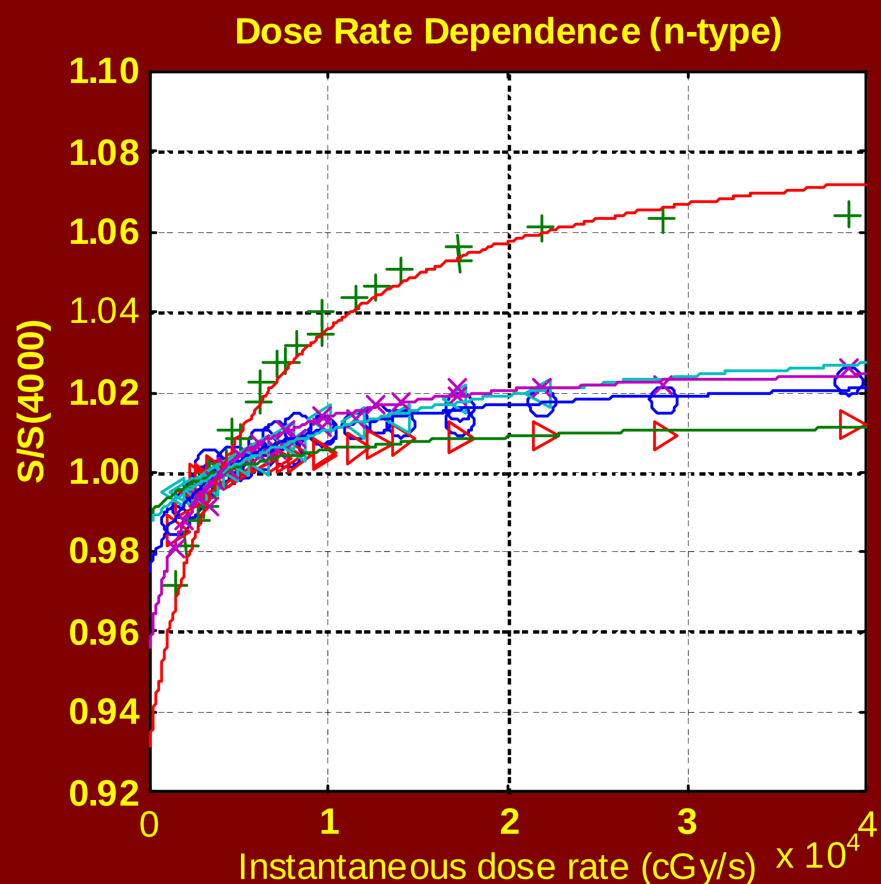
$\boxed{CF = D_0(Q)\ /\mathrm{Signal}(D_0,Q)}$

$\boxed{\mathrm{cGy/Signal}}$

SSD Dependence (Ratio at 200 cm)

Diode	6 MV	20 MV	Co-60
Isorad 1 Gold	0.950	0.957	0.988
Isorad 2 Gold	0.965	0.973	---
Isorad Red	0.974	0.994	0.987
SPD	1.002	0.995	---
EDP30	0.995	0.998	0.994

Dose Rate Dependence

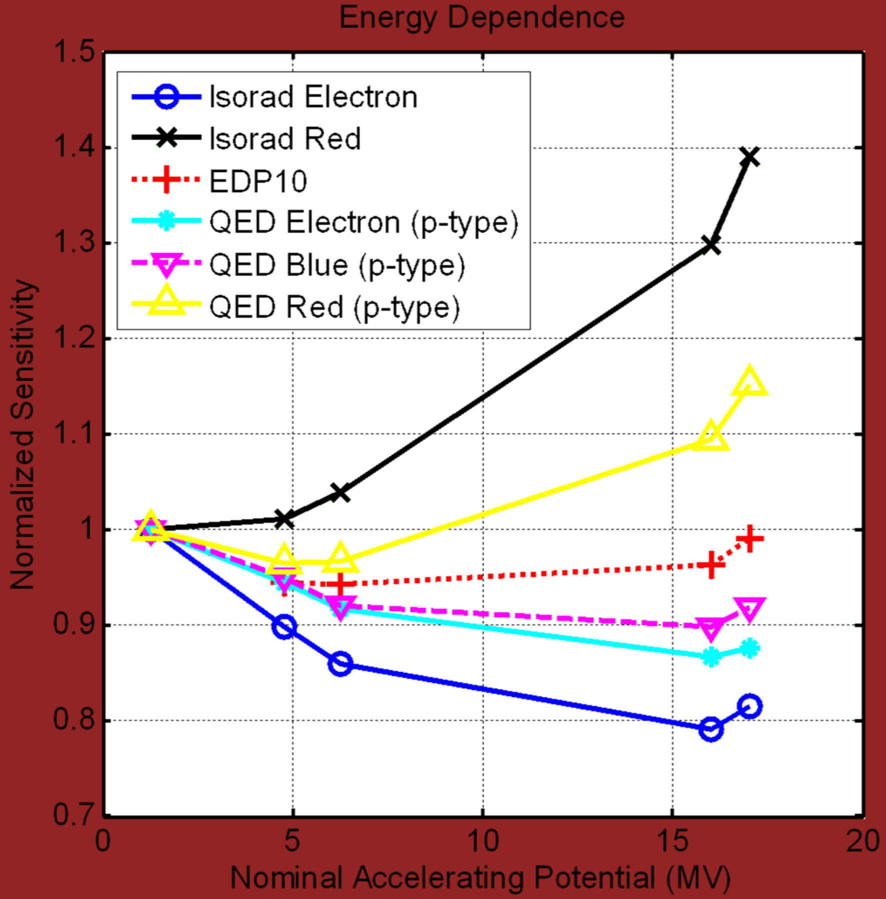


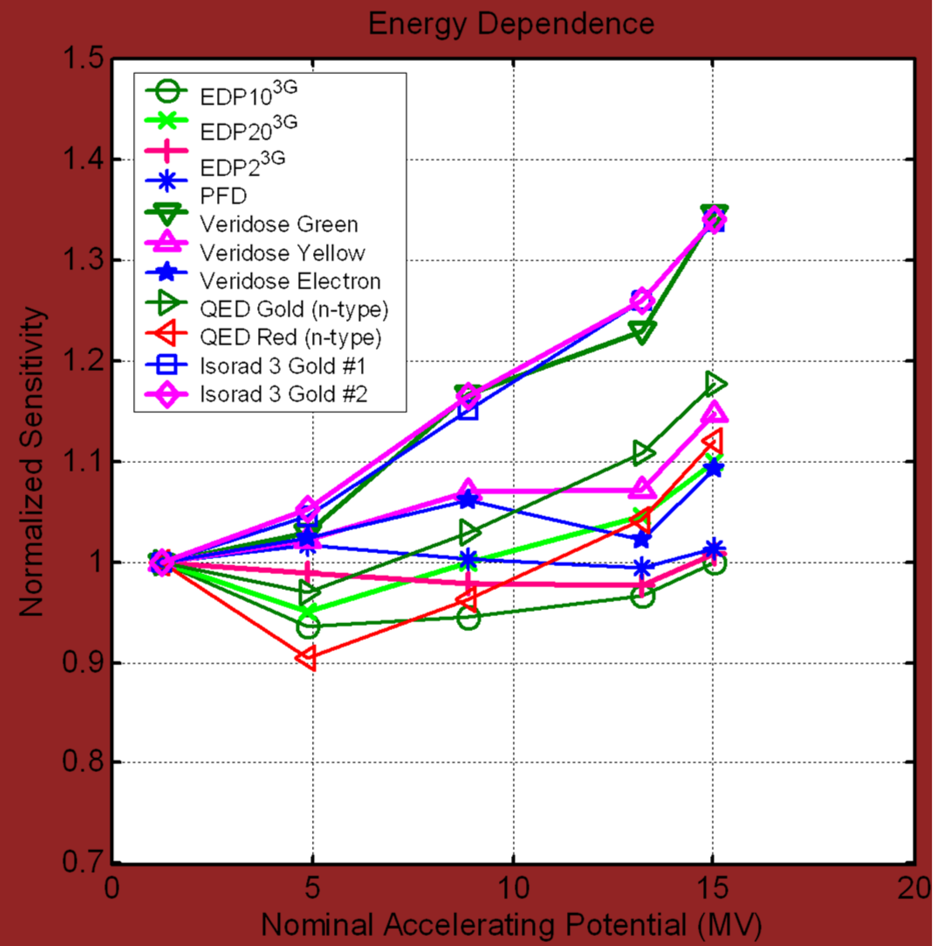
o-Isorad Gold#1 + - Isorad Red (n-type), ▹ - Isorad-3 Gold, ◃ - Veridose Green x - QED Red (n-type)

◊ - EDP103G , x- EDP203G , * - Isorad-p Red, Δ - QED Red (p-type), ∇- QED Blue

(Saini and Zhu, 2004)

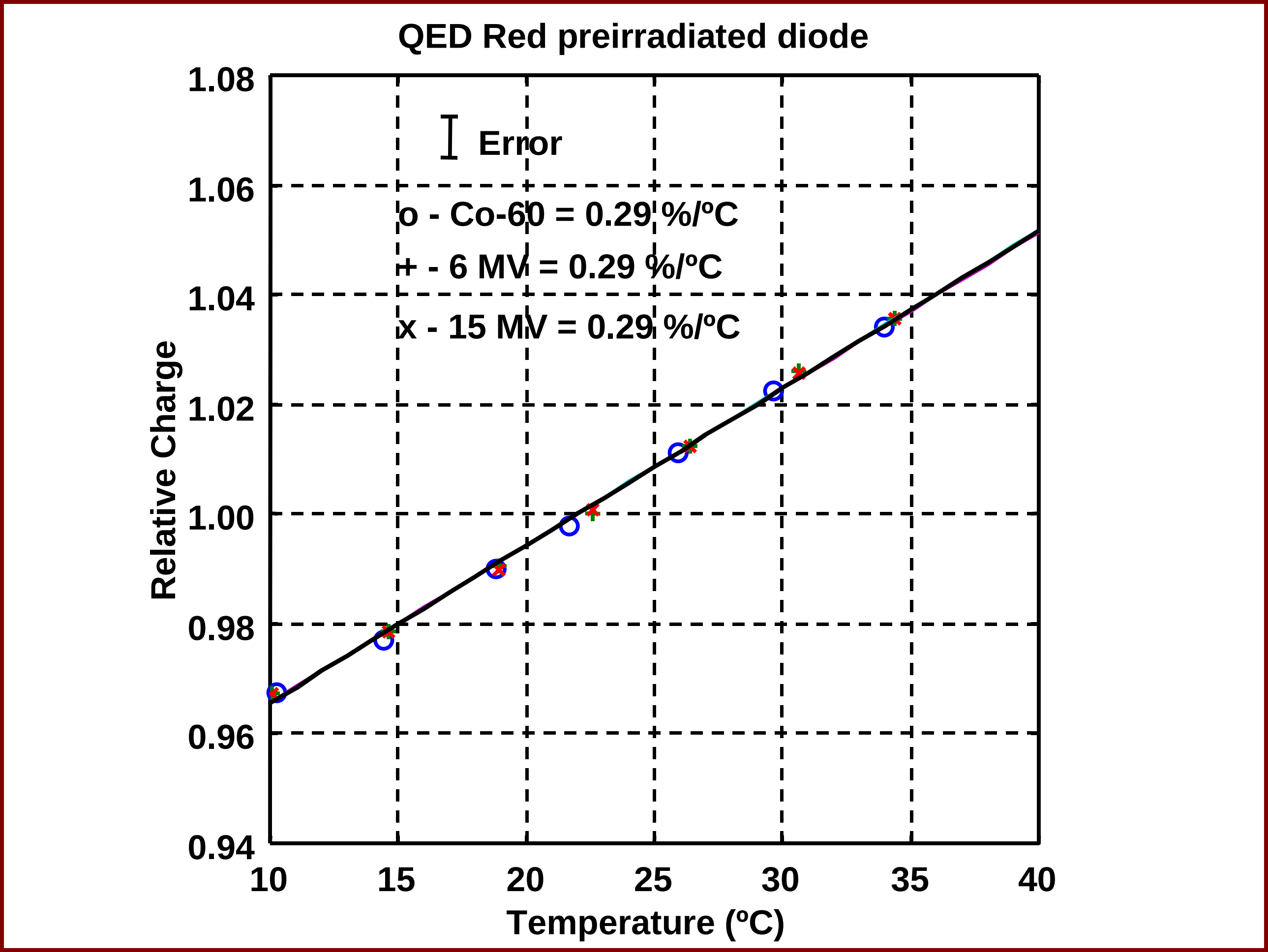
Energy Dependence for megavoltage photon





(Saini and Zhu, 2007)

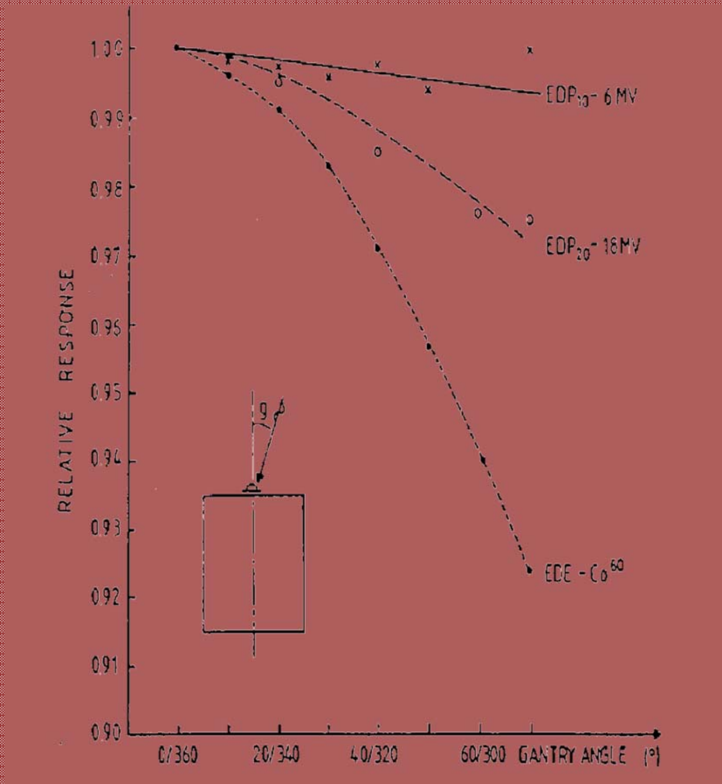
Temperature coefficient
≈ 0.3% / ^{0}C



(Saini and Zhu, 2002)

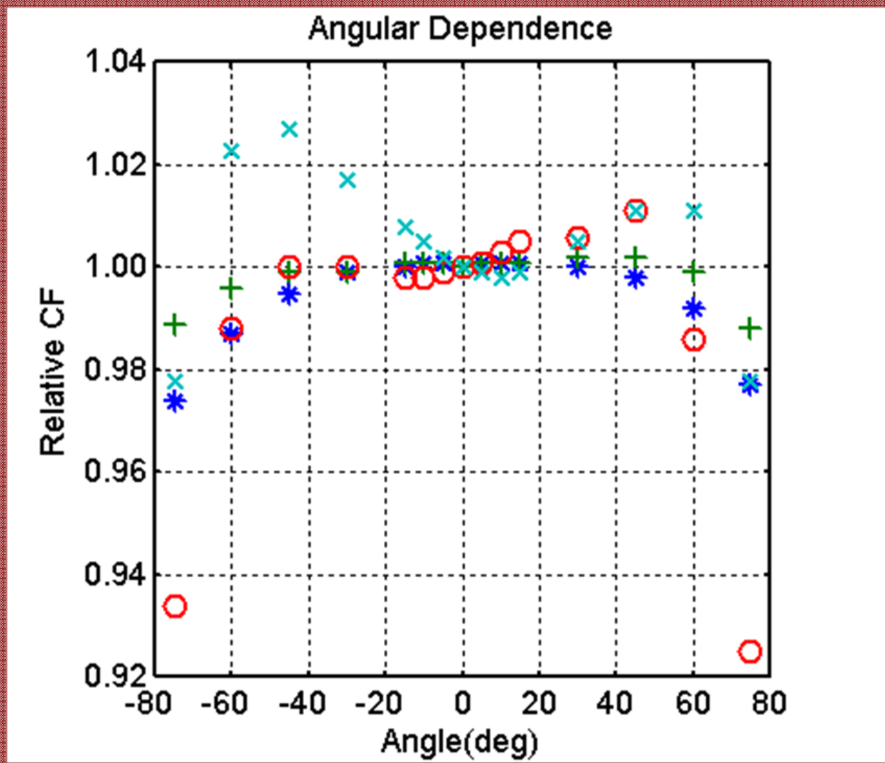
Angular Dependence

Scanditronix



(Rikner, Thesis, 1983)

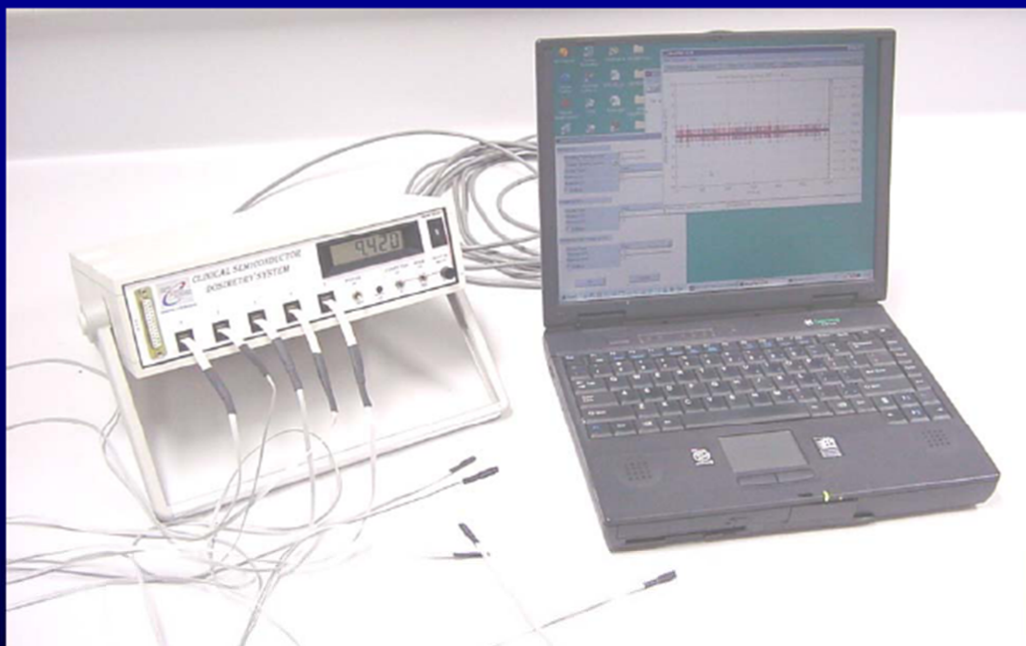
Sun Nuclear



* - Isorad-3 Gold, + - Isorad-3 Red, O - QED Gold, x - QED Red

(Saini, Thesis, 2007)

CMRP MOSFET Dosimetry System



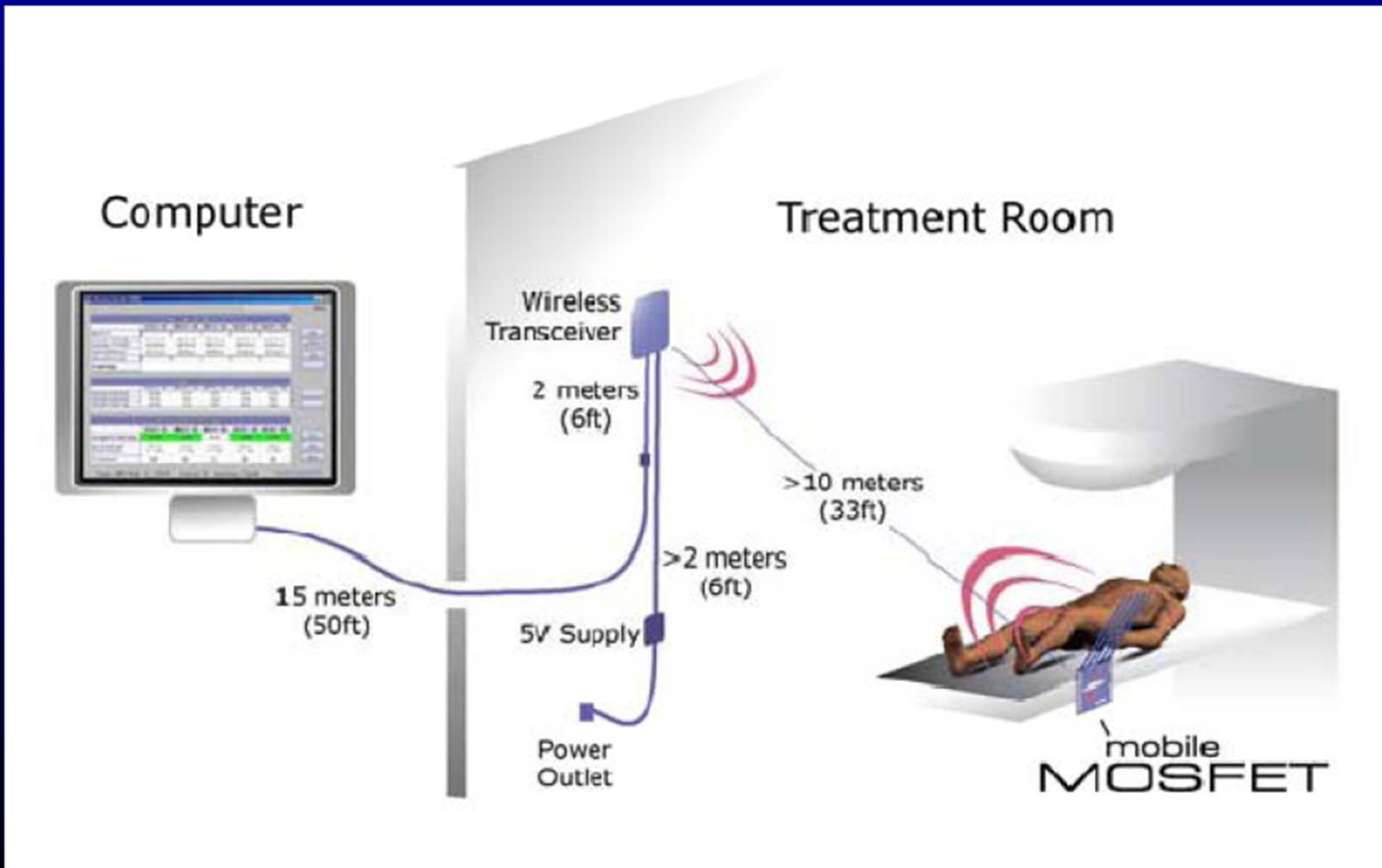
MOSkin detectors, thickness 0.07 mm, see Table 29-III

MOSFET Clinical Dosimetry System:

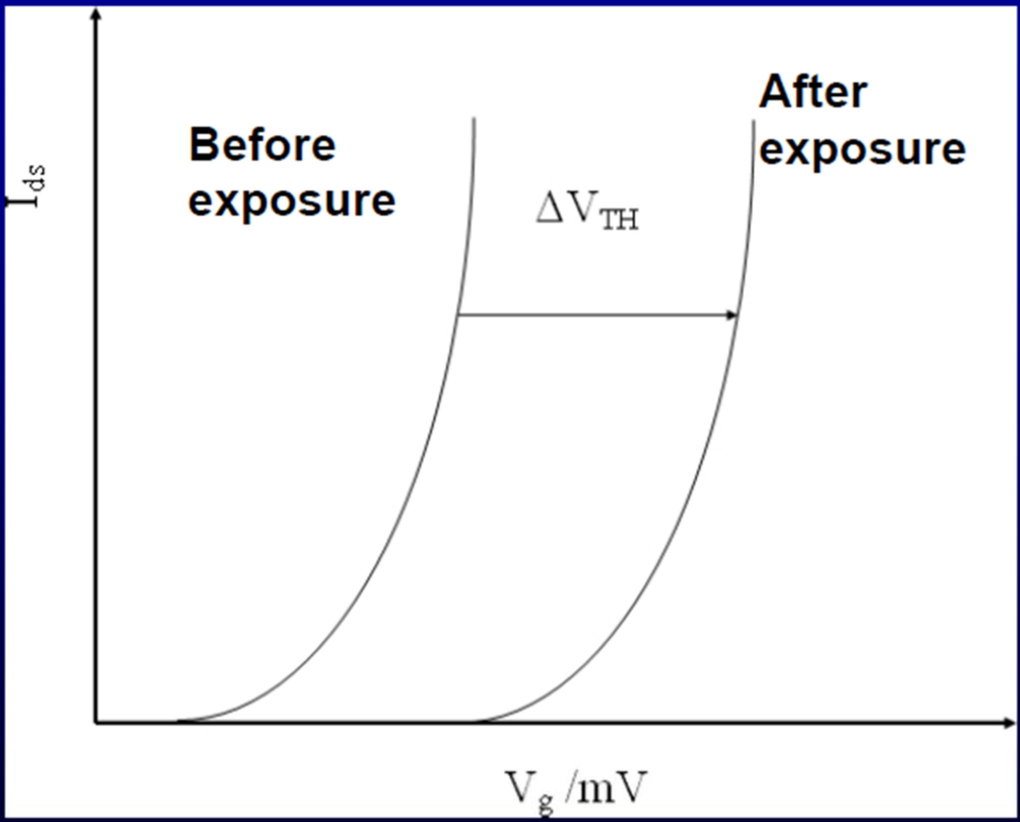
designed and distributed by CMRP

Courtesy of Anatoly Rosenfeld

Wireless setup



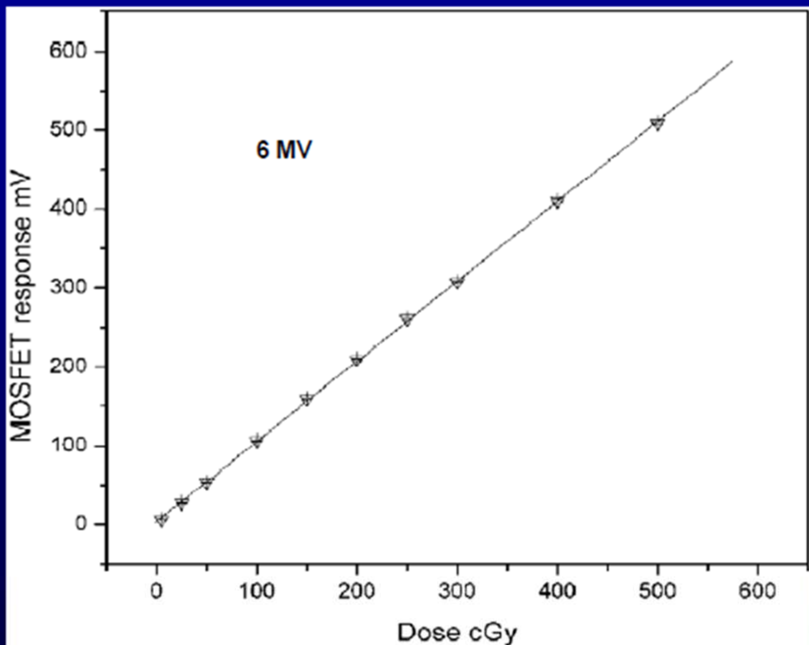
Threshold voltage shift / ΔVT

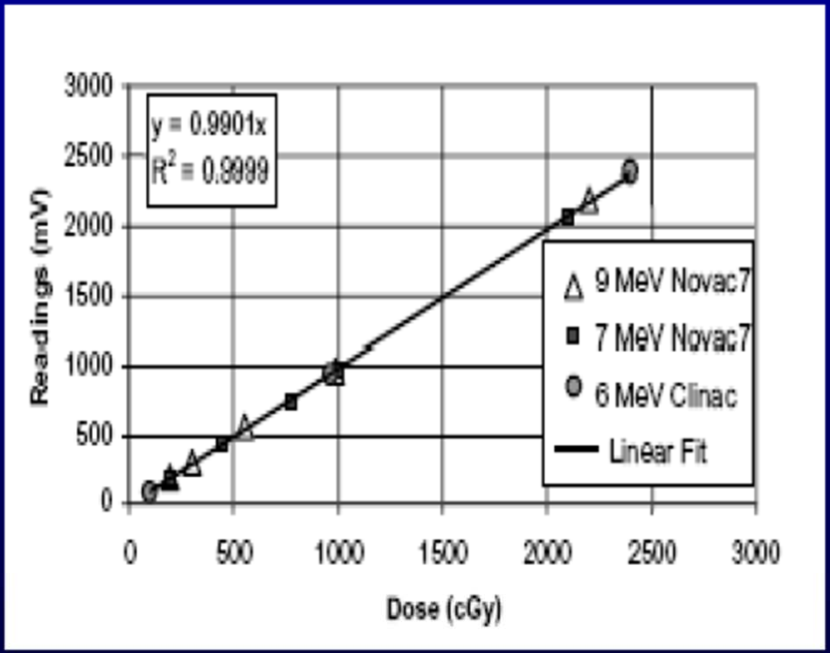


- ΔV_T is a function of absorbed dose
- That function is linear when the MOSFET operates in the biased mode during the irradiation
- Absorbed dose linearity region increases with the increase of the bias voltage

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

Absorbed dose linearity – dual MOSFET-dual-bias

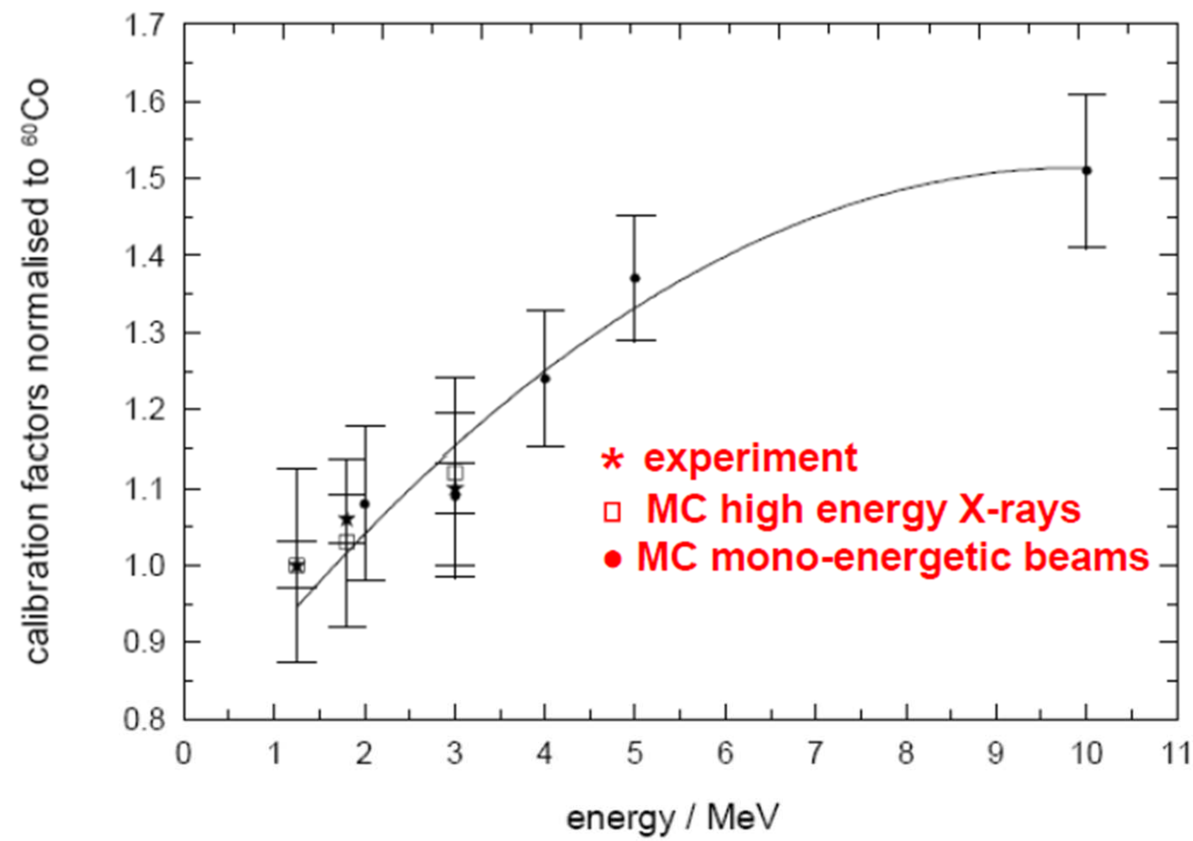




Ramaseshan et al, Phys. Med. Biol. 49 , 4031-4048, 2004

Consorti et al, Int. J. rad. Onc. Biol. Phys. 63, 952-60, 2005

Energy dependence - TN-502-RD

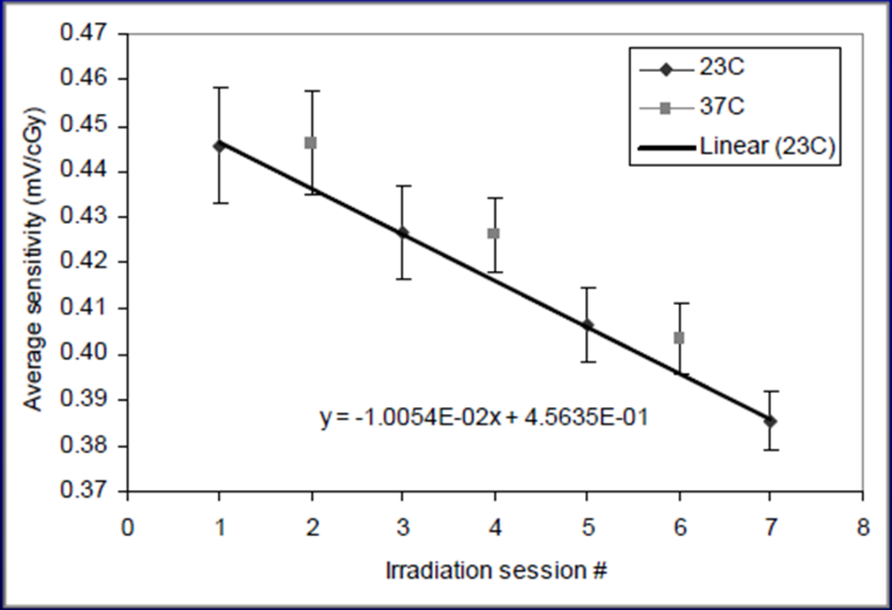


Panettieri et al Phys Med Biol. 52(1):303-16.2007

Temperature dependence

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

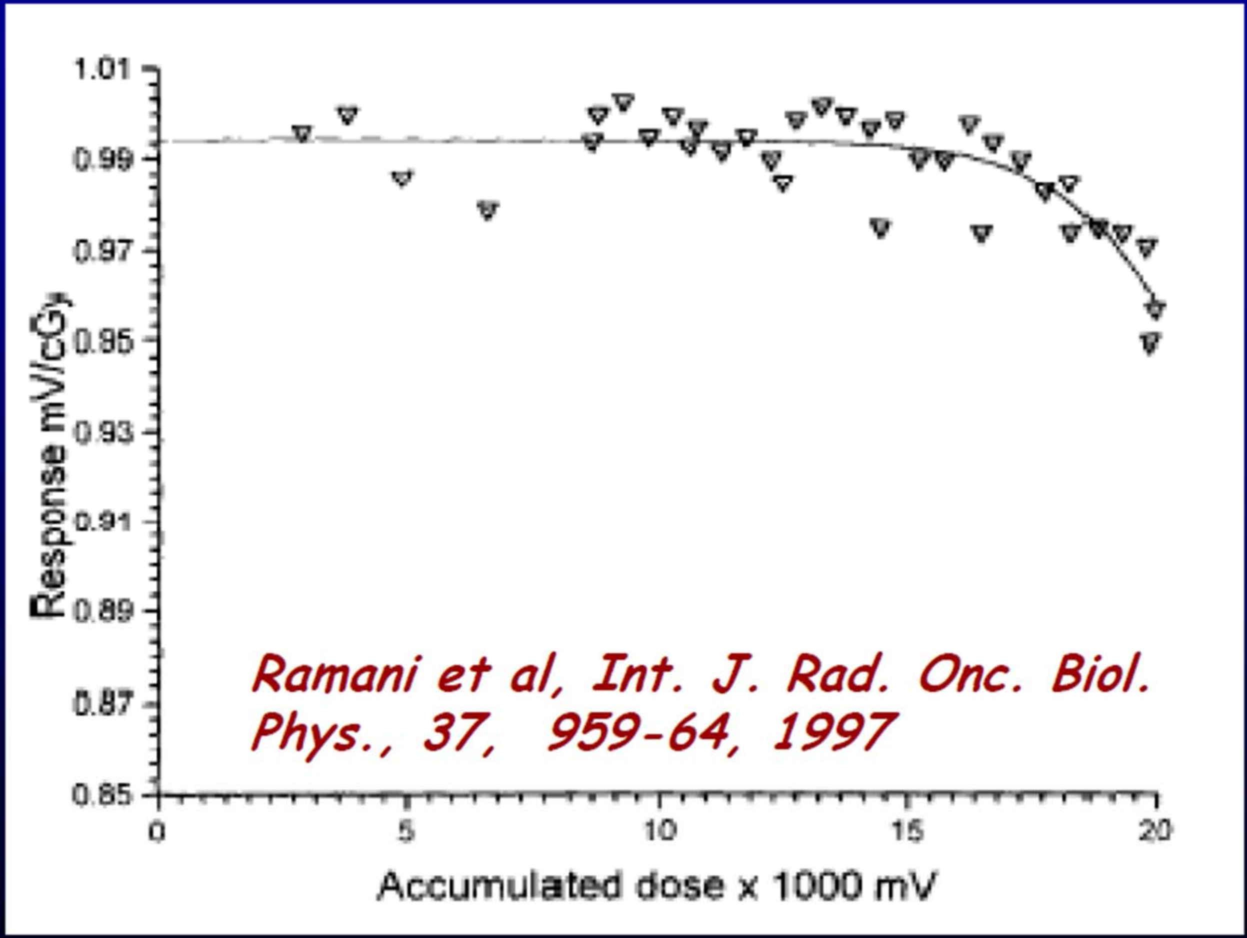
DVS dosimeter is approximately 3.3% more sensitive (higher dose reading for same applied dose) when irradiated at 37°C vs. 23°C



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

Passive In Vivo Dosimeters

Luminescent In Vivo Dosimeters
TLDs and OSLDs
TG-191

Recommendations on the clinical use of luminescent detectors

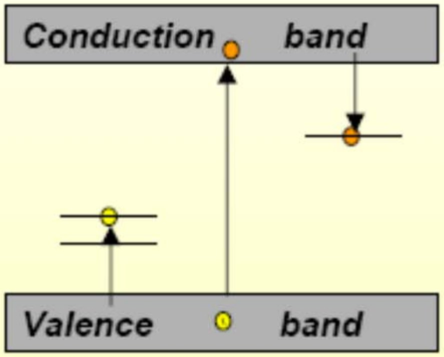
Eduardo G. Yukihara
and
Stephen. W.S. McKeever

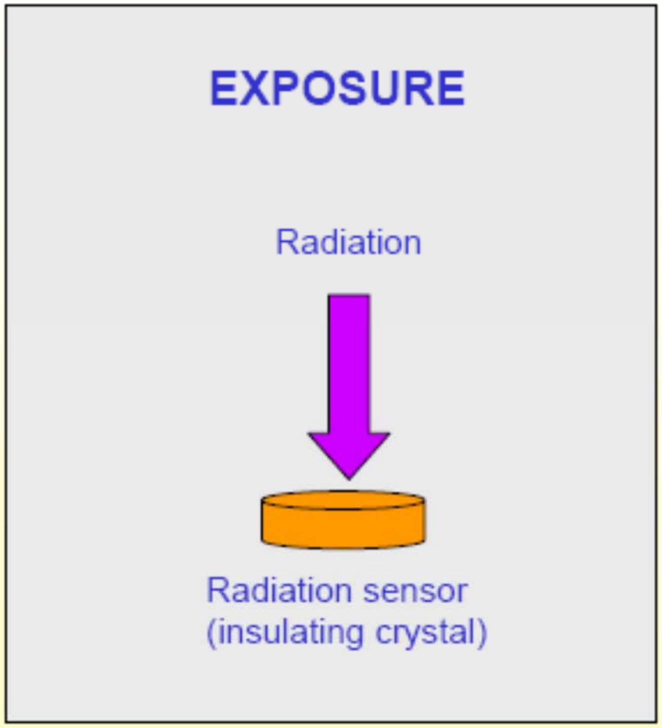
Department of Physics,
Oklahoma State University

TL and OSL Principles

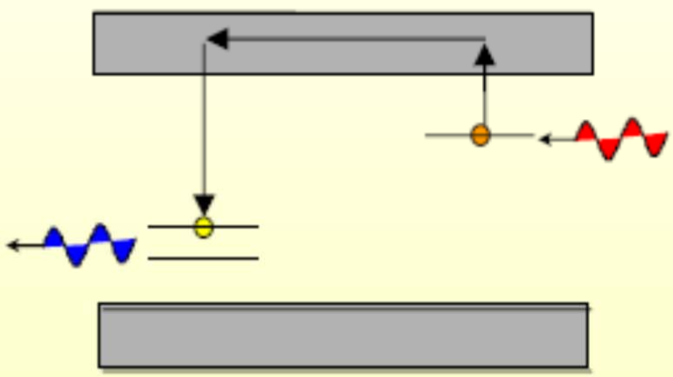
TL = *thermally* stimulated luminescence

OSL = *optically* stimulated luminescence





TL and OSL Principles



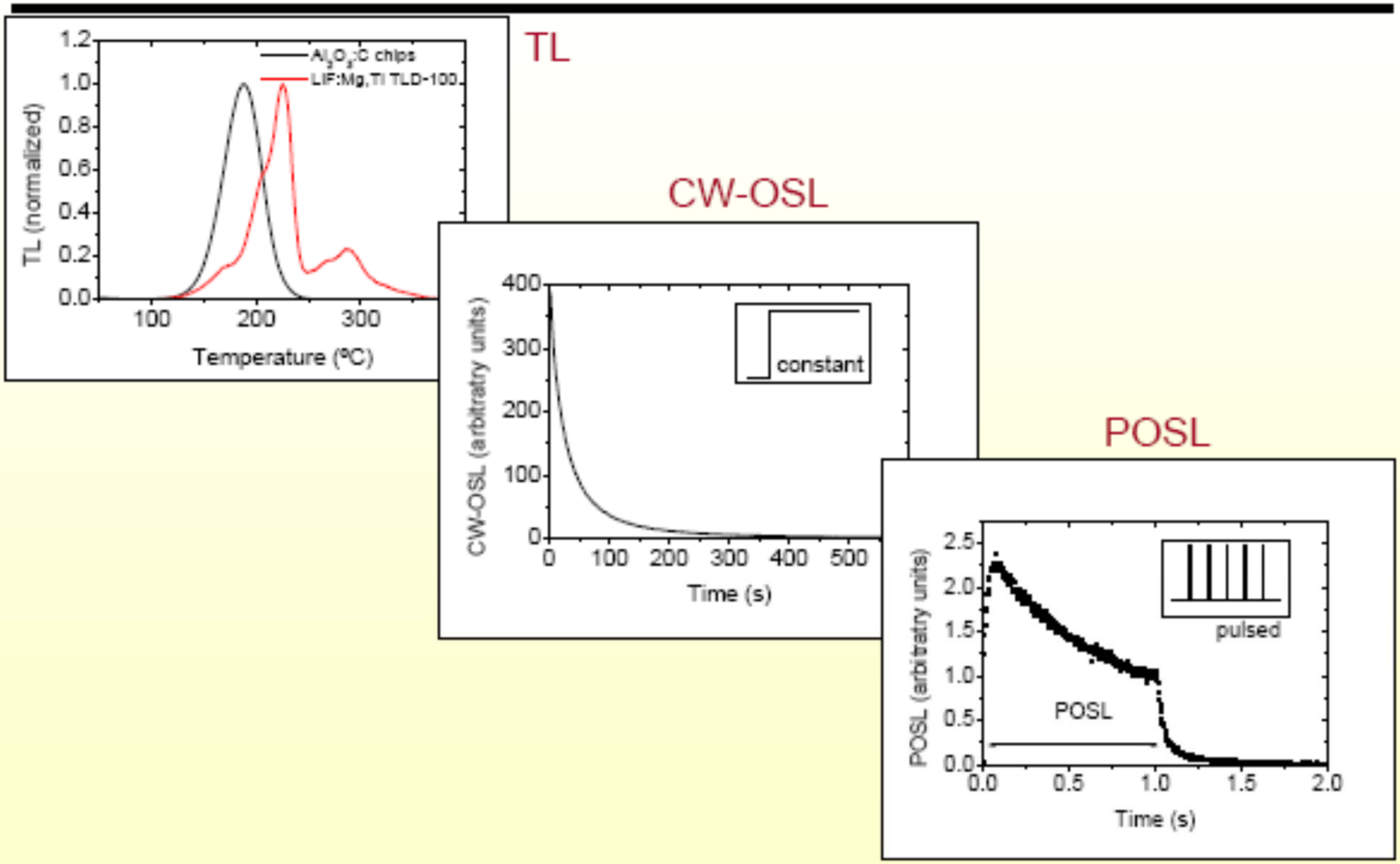
READOUT

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

Thermal
stimulation
(heating)

Yukihara and McKeever

Typical TL and OSL Signals

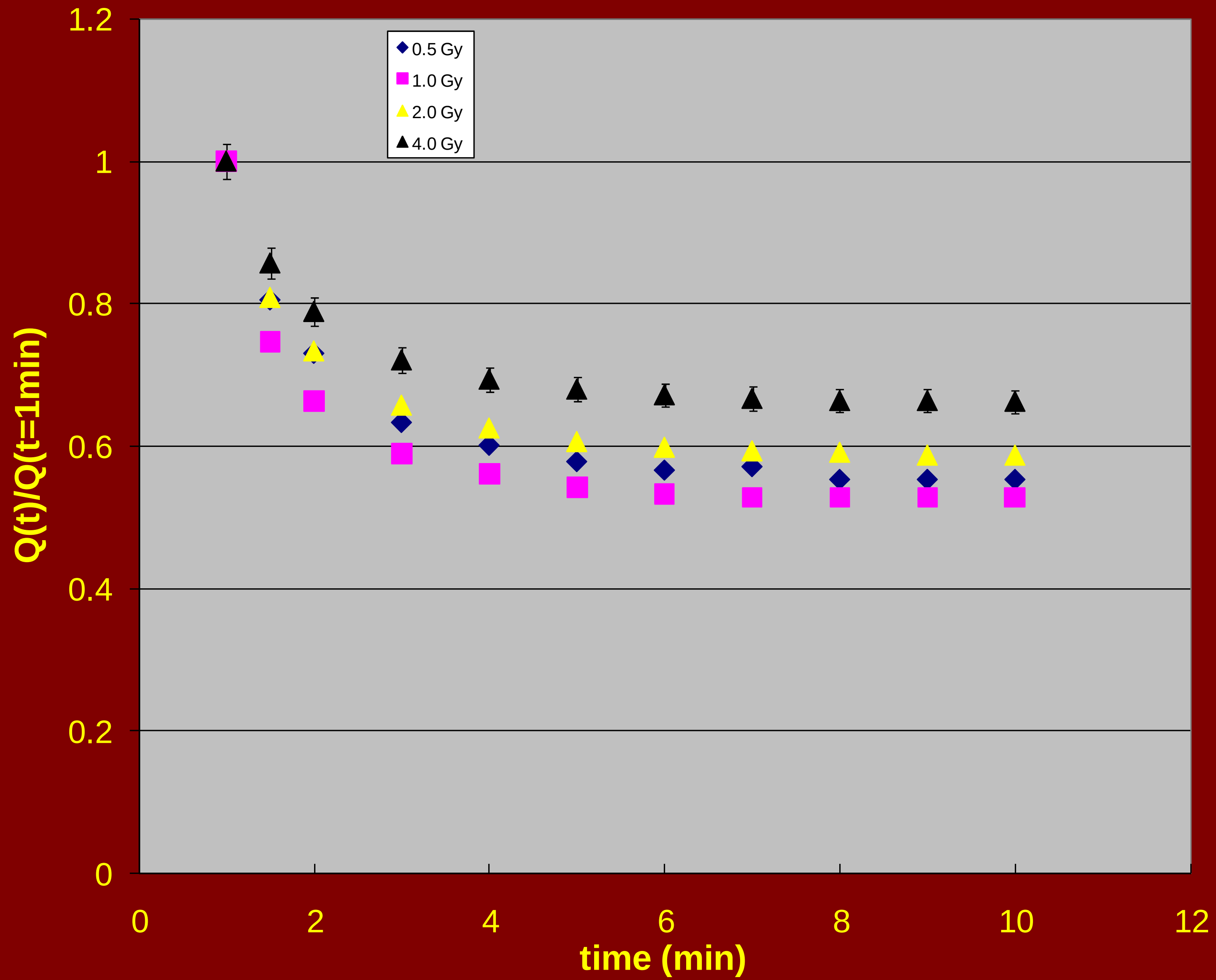


Oklahoma State University

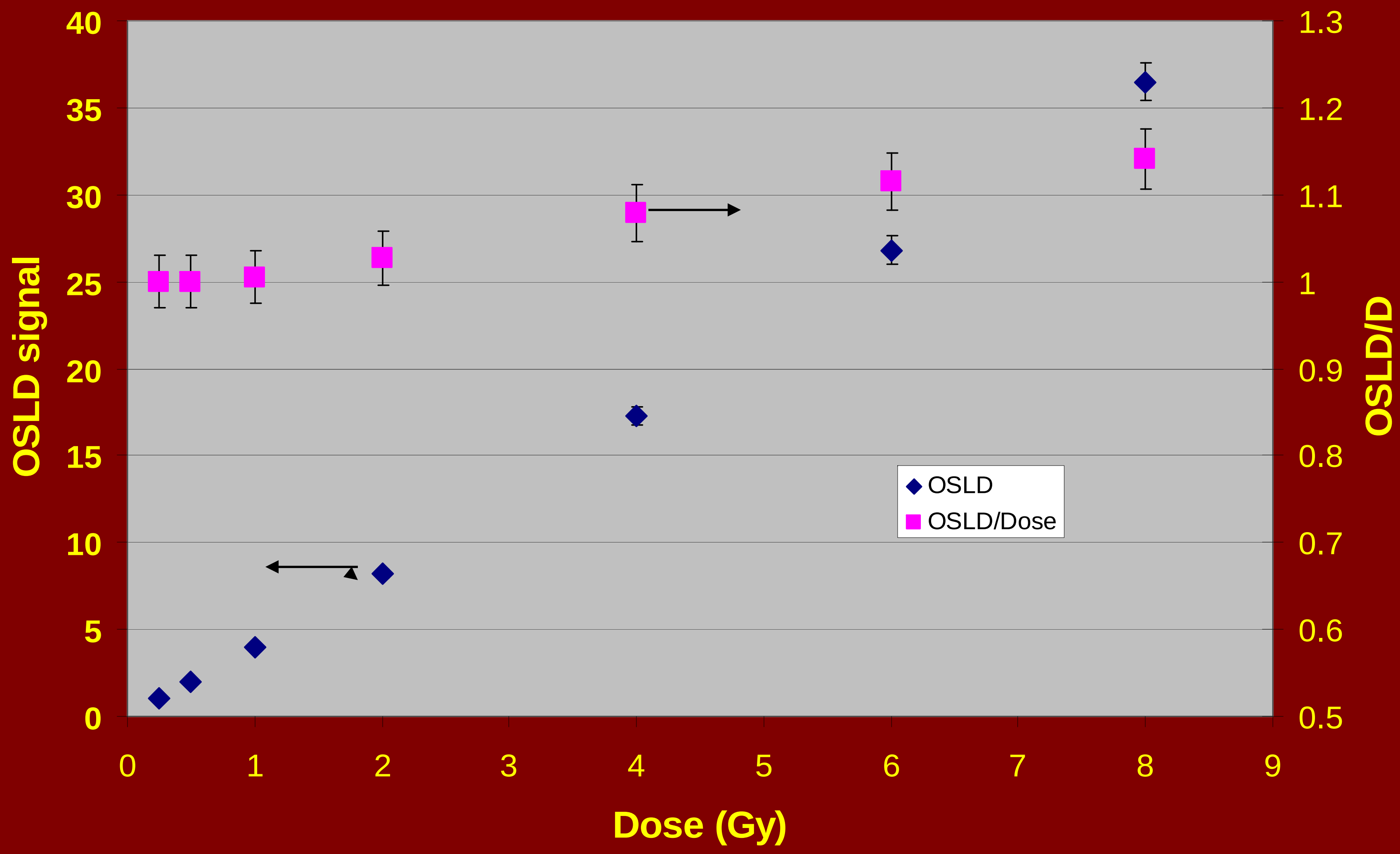
Yukihara and McKeever

microStar
NITROGEN OUTLET
lenovo
InLight
Loader
LED
ON
Indicator
LANDAUER®
Measuring
Position
Dial
H/P
DRK
CAL
LED
Power
ON
Indicator
microStar

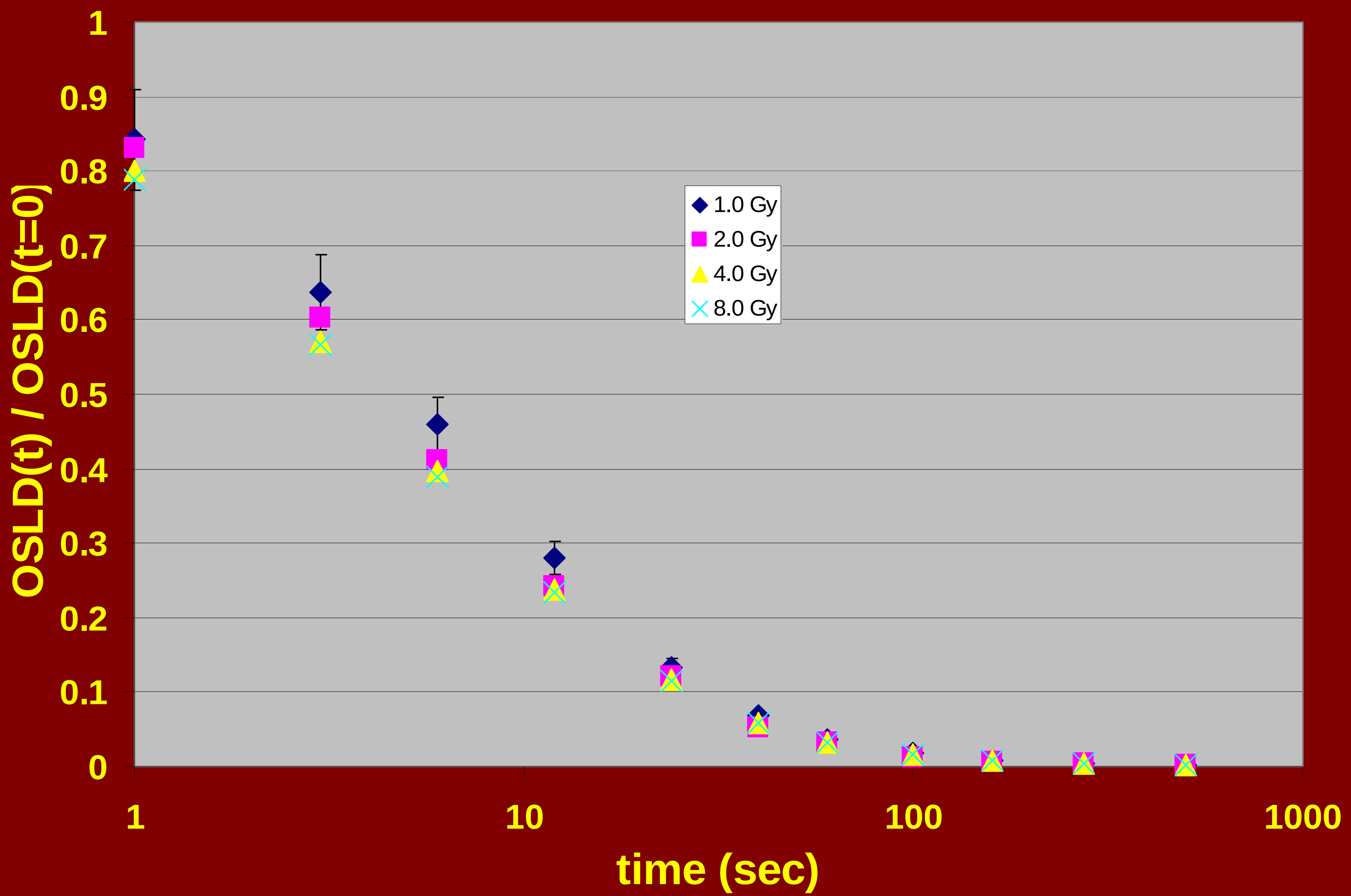
Fig. 3 Decay of DOT signal with time at room temperature



OSLD and OSLD/Dose as a function of dose normalized to their readings at 0.25 Gy



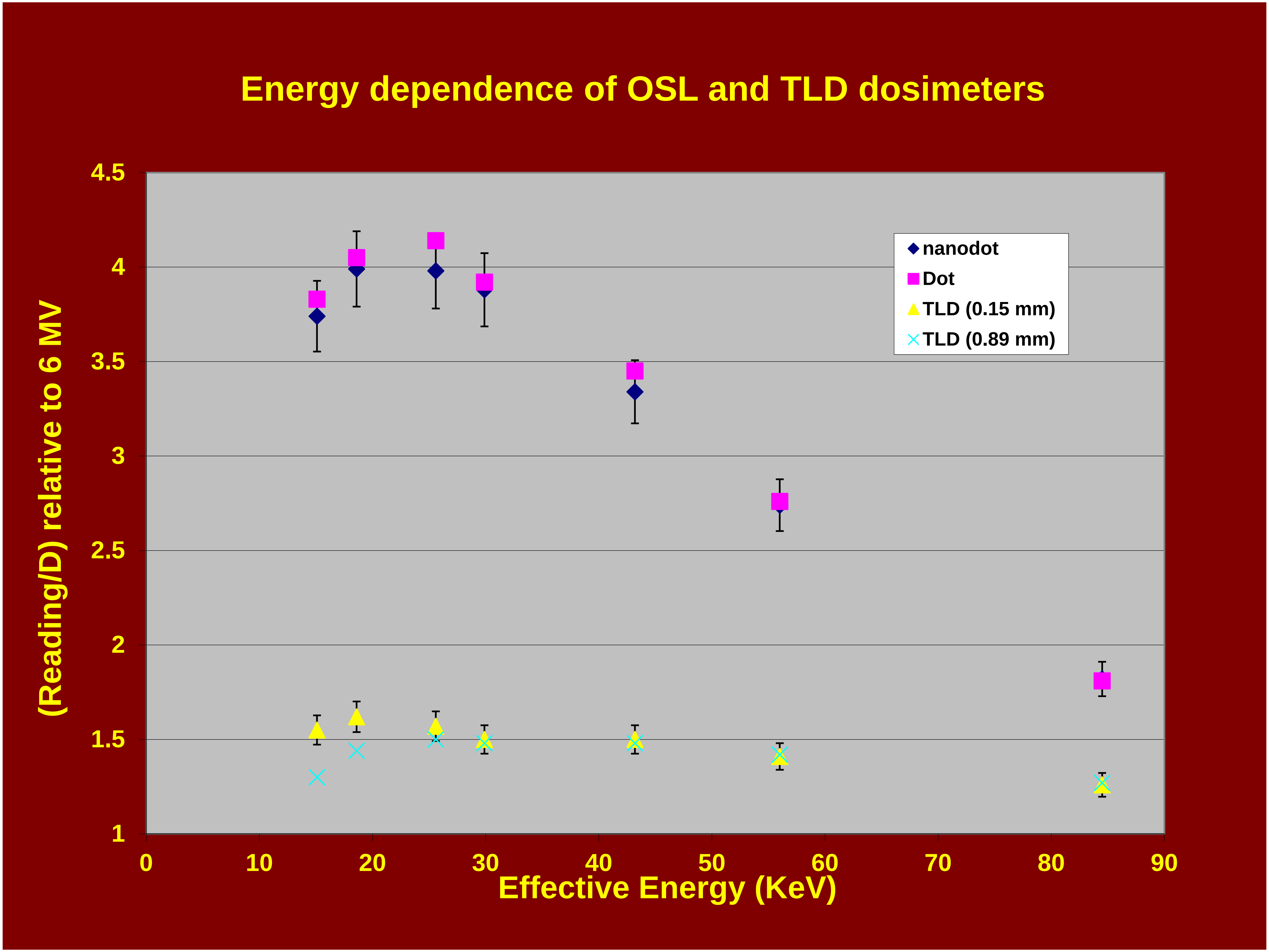
Optical bleaching with 15 W lamp of OSLDs irradiated to various doses



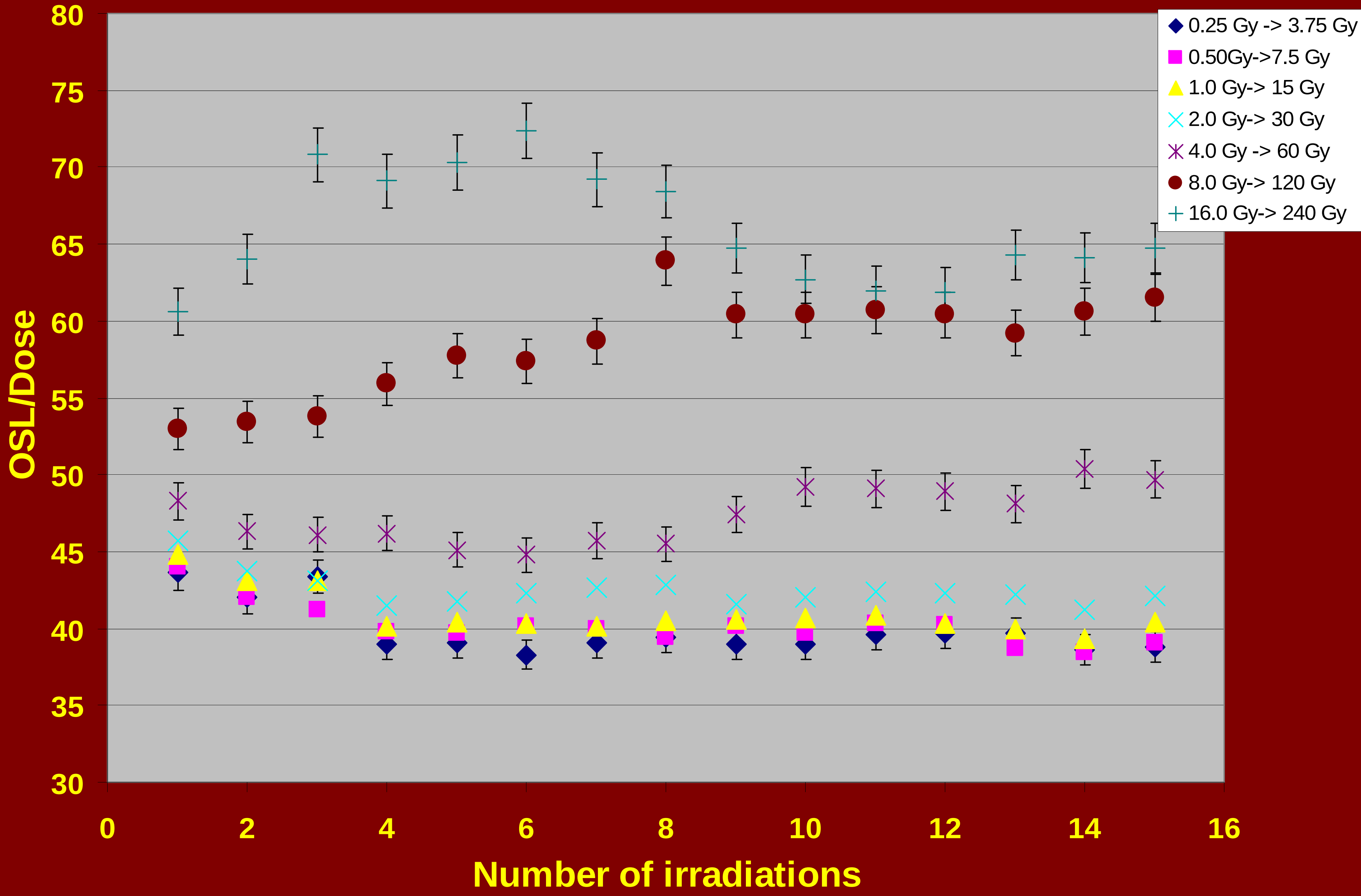
Energy response of TLD and OSLD detectors normalized to their response at 6 MV

Modality	Energy MV	Equivalent Energy MeV	$(TLD/D)^{Q}_{6MV}$	$(OSLD/D)^{Q}_{6MV}$
photons	0.250	0.110	1.28	1.71
"		1.25	1.02	1.04
"	6	2.4	1.00	1.00
"	18	5.7	0.99	0.99
electrons		6	0.98	0.99
"		9	0.99	0.99
"		12	0.99	0.98
"		16	0.99	0.98
"		20	0.99	0.98
protons		100	1.08	1.05
"		180	1.06	1.06
"		250	1.08	1.07

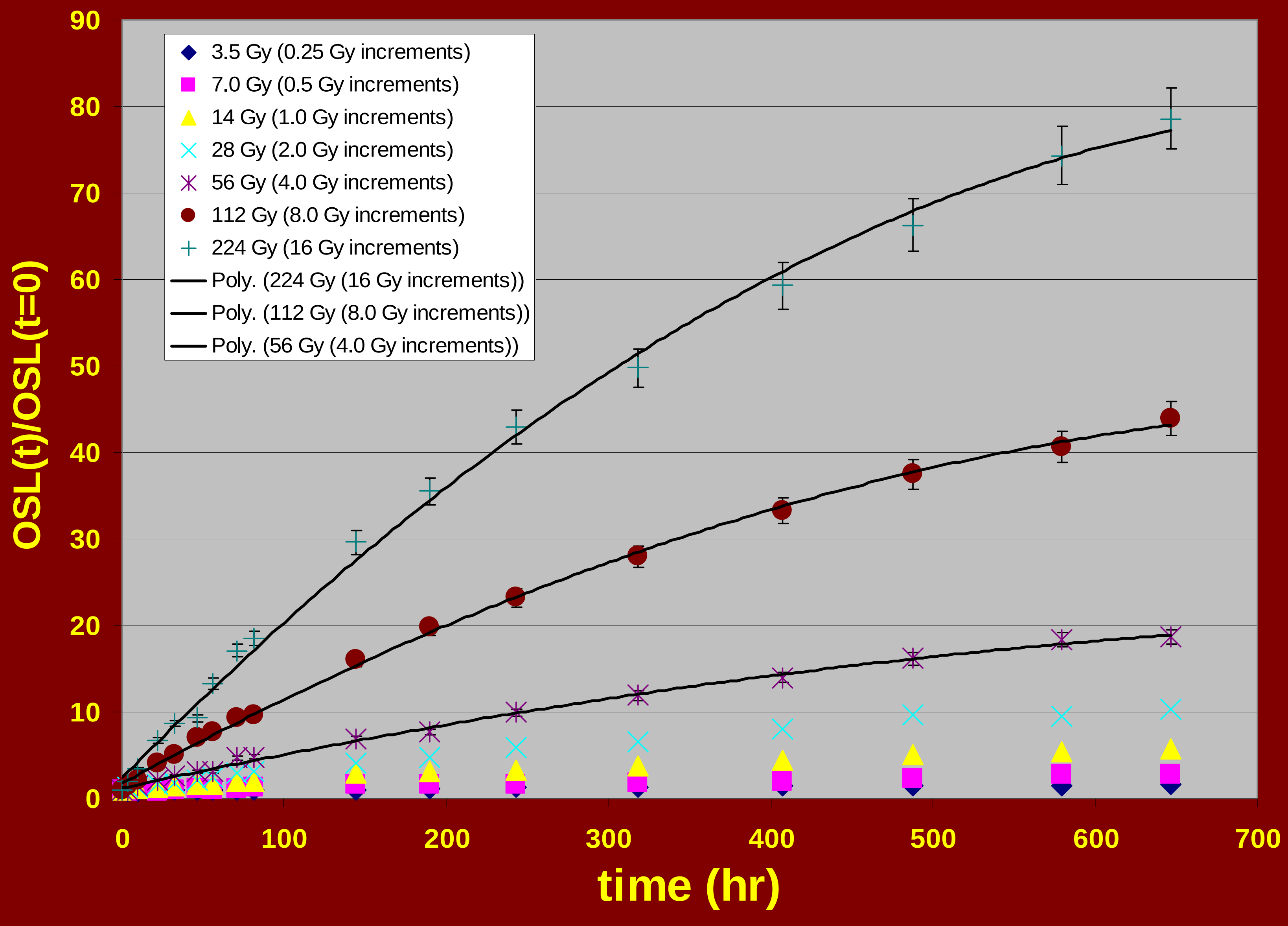
$\sigma \approx \pm 5.5\%$, and $\pm 3.8\%$ within 1 SD for kilovoltage and megavoltage irradiations, respectively


Energy dependence of OSL and TLD dosimeters
(Reading/D) relative to 6 MV
4.5
4
3.5
3
2.5
2
1.5
1
0
10
20
30
40
50
60
70
80
90
Effective Energy (KeV)
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.



Increase in background signal with absorbed dose



Advantages of TLDs

- Wide useful dose range mrad → 10^3 rad (linearity)
- Dose-rate independence 0 → 1011 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

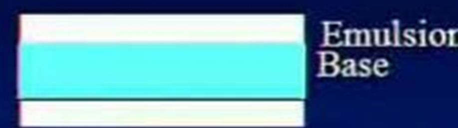
- **encapsulated in light tight plastic housing**
- **optical bleaching cannot clear all the radiation effects resulting in an increased background signal**
- **sensitivity changes with accumulated doses > 20 Gy**

Film Dosimetry

- Radiographic
- Radiochromic

Radiographic Film

- Base (Cellulose nitrate or Polyester) (typically 200 μm)

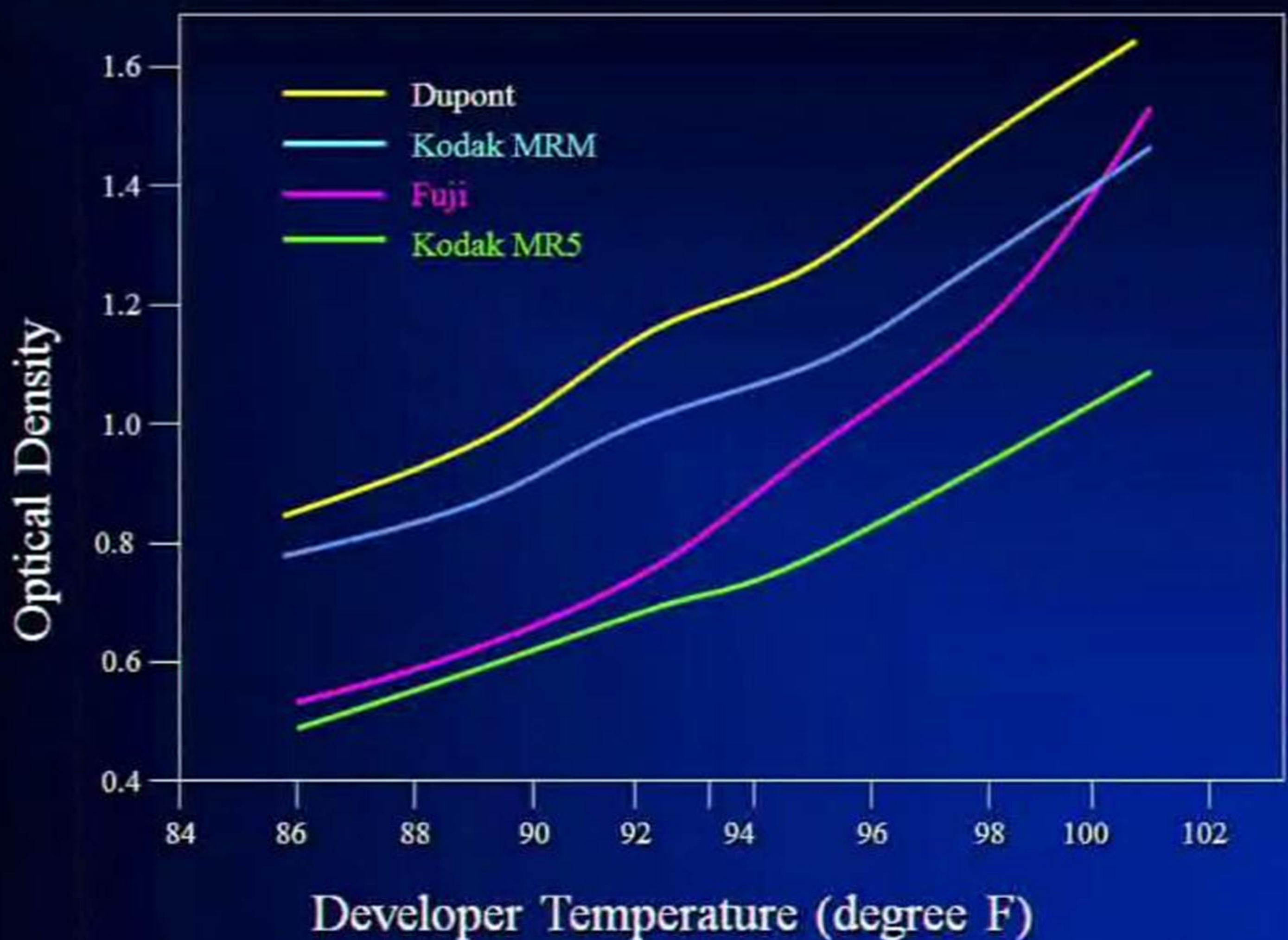


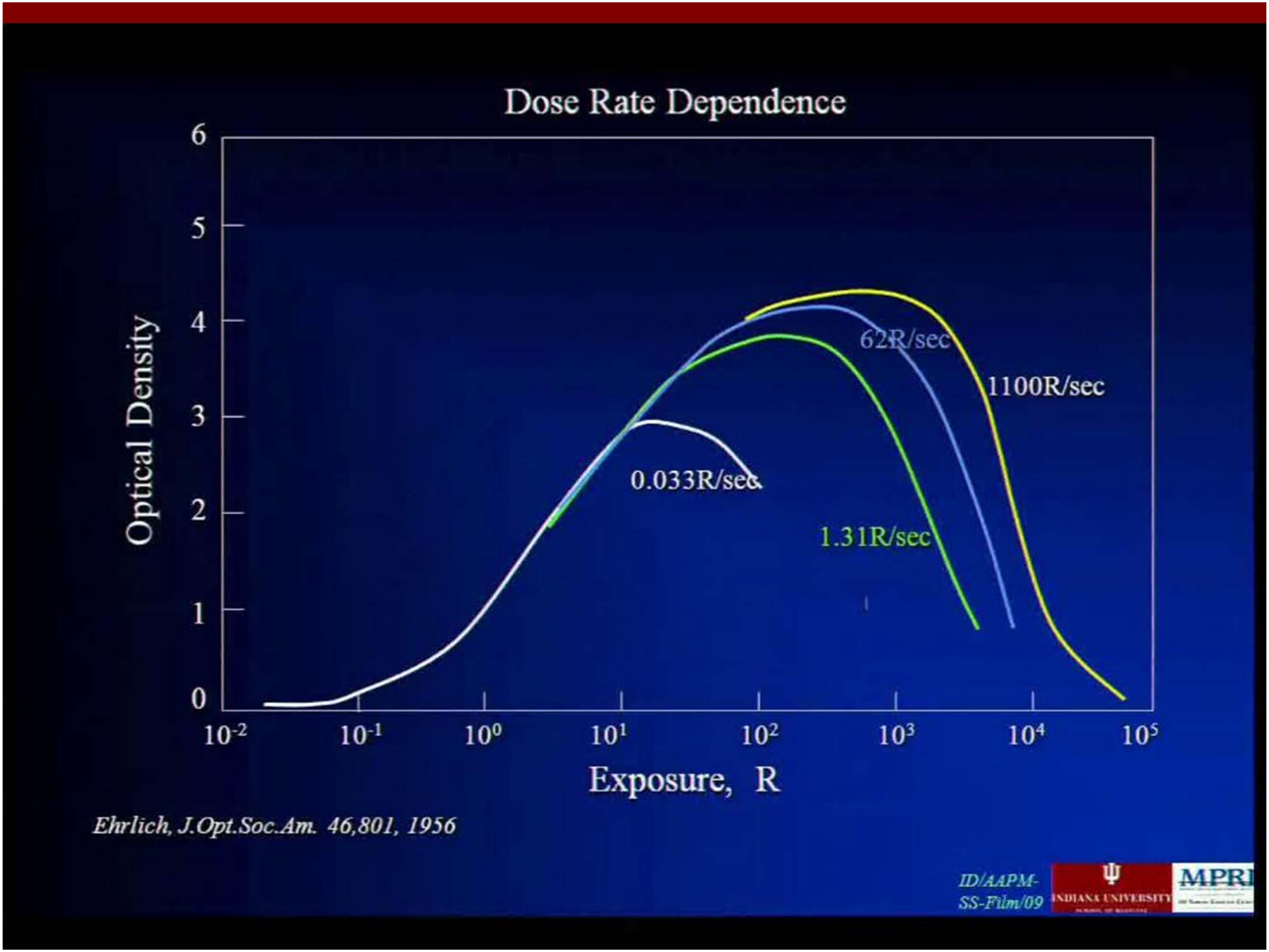
- Emulsion (10-20 μm; 2-5 mg/cm^3)
  - Gelatin (derivative from bone)
  - grain (size: 0.1~3 μm diameter)
    - AgBr (cubic crystal with lattice distance of 28 nm
    - AgI
    - KI
      - There are 10^9-10^{12} grains/cm^2 in a x-ray films
- Coating
  - Very sensitive which may determine X & Y direction 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

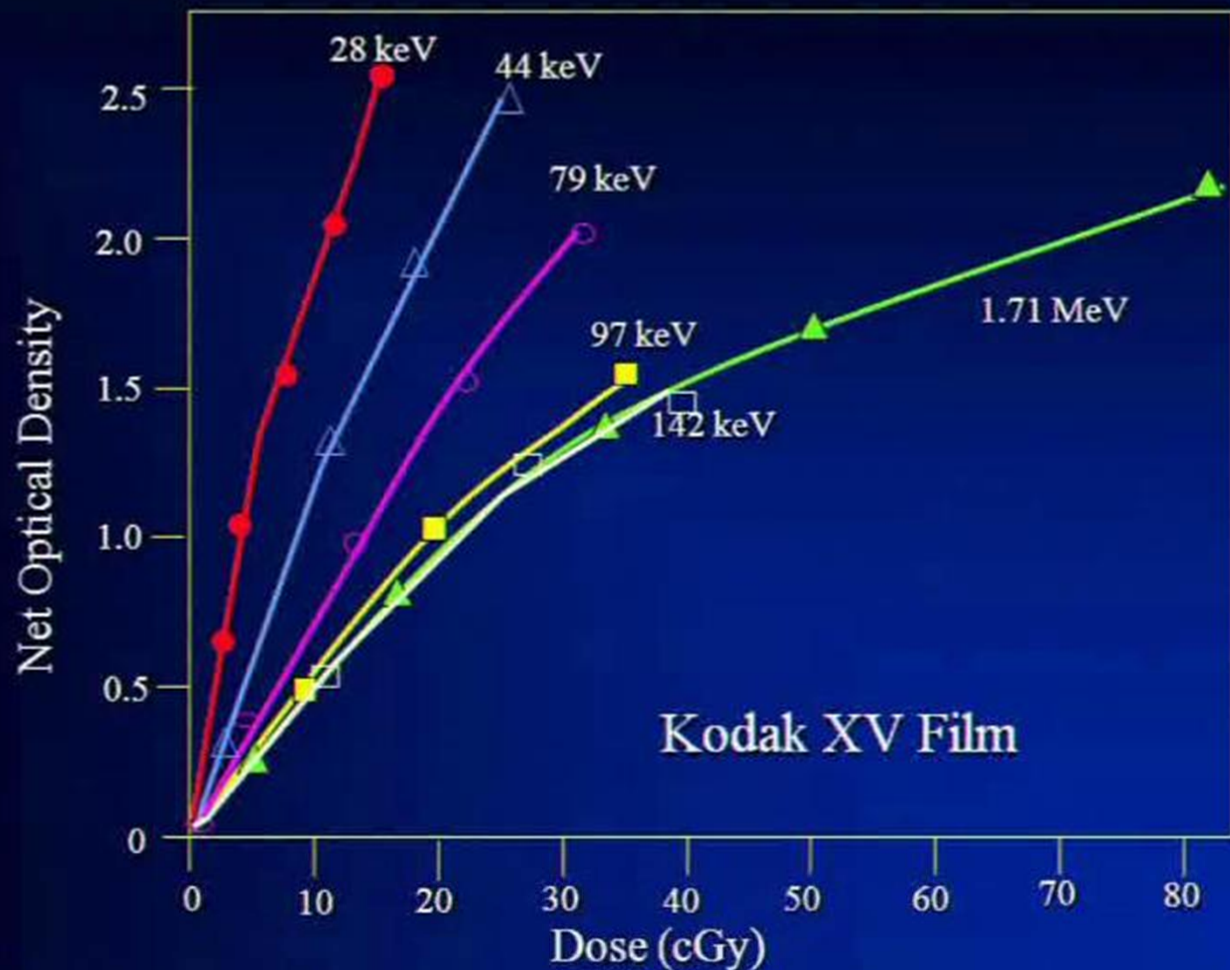
Temperature Dependence of Various Films





Dose Rate Dependence
Optical Density
6
5
4
3
2
1
0
10^{-2}
10^{-1}
10^{0}
10^{1}
10^{2}
10^{3}
10^{4}
10^{5}
Exposure, R
62R/sec
1100R/sec
0.033R/sec
1.31R/sec
Ehrlich, J.Opt.Soc.Am. 46,801, 1956
ID/AAPM-SS-Film/09
INDIANA UNIVERSITY
MPRI

Energy Dependence of Radiographic Film



Muench et al, Med. Phys. 18, 769, 1991

INDIANA UNIVERSITY
MPRI

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

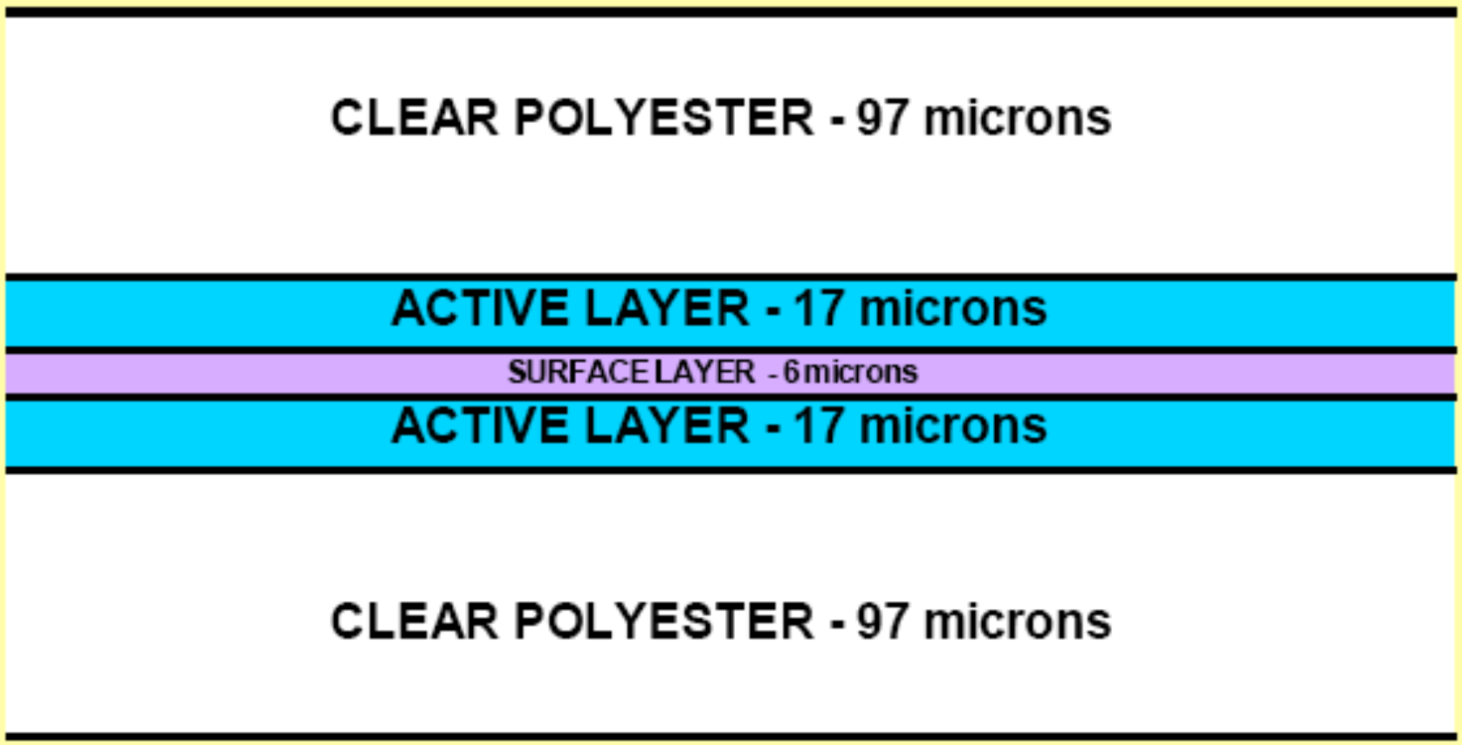


Figure 1: Configuration of GAFCHROMIC® EBT Dosimetry Film

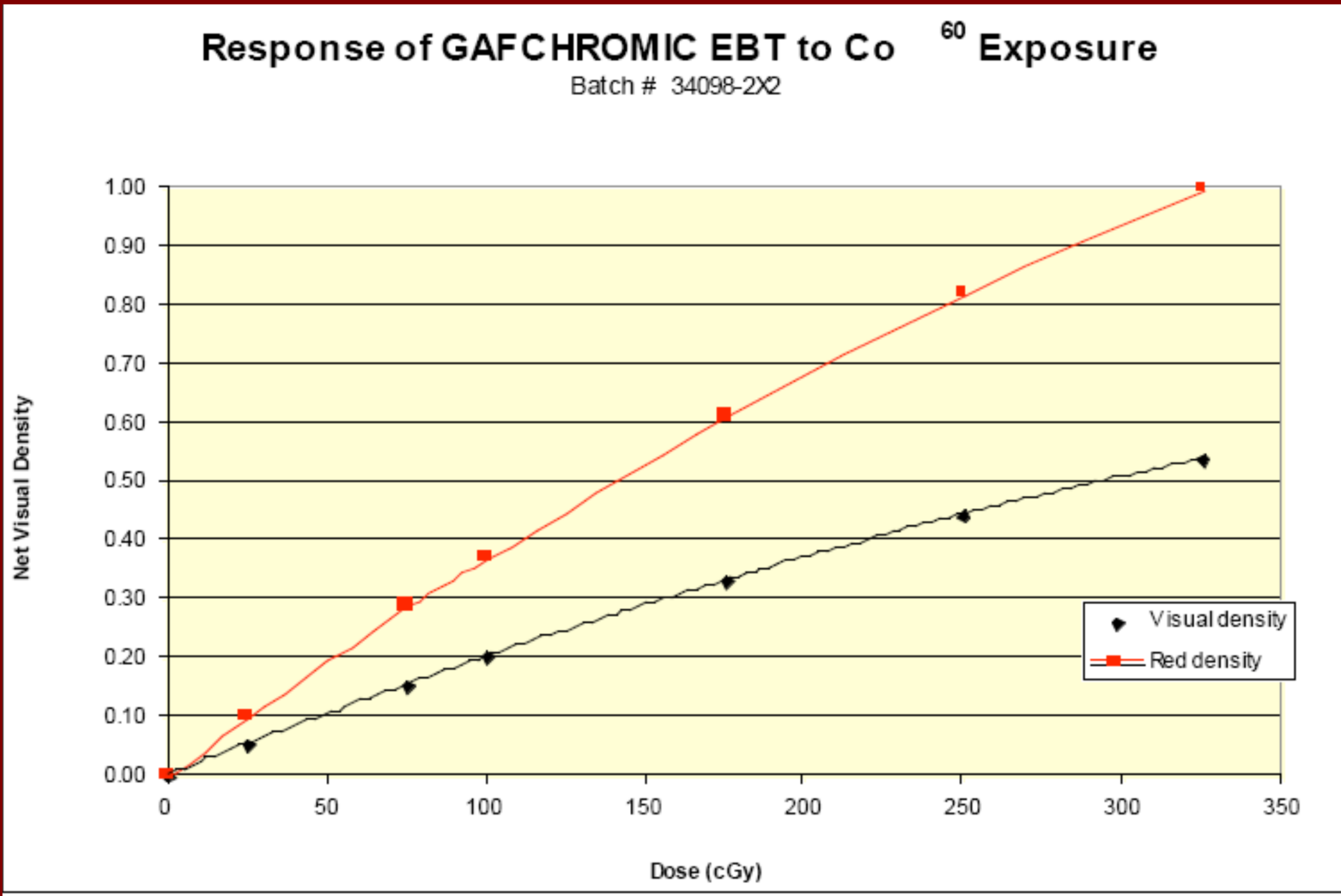
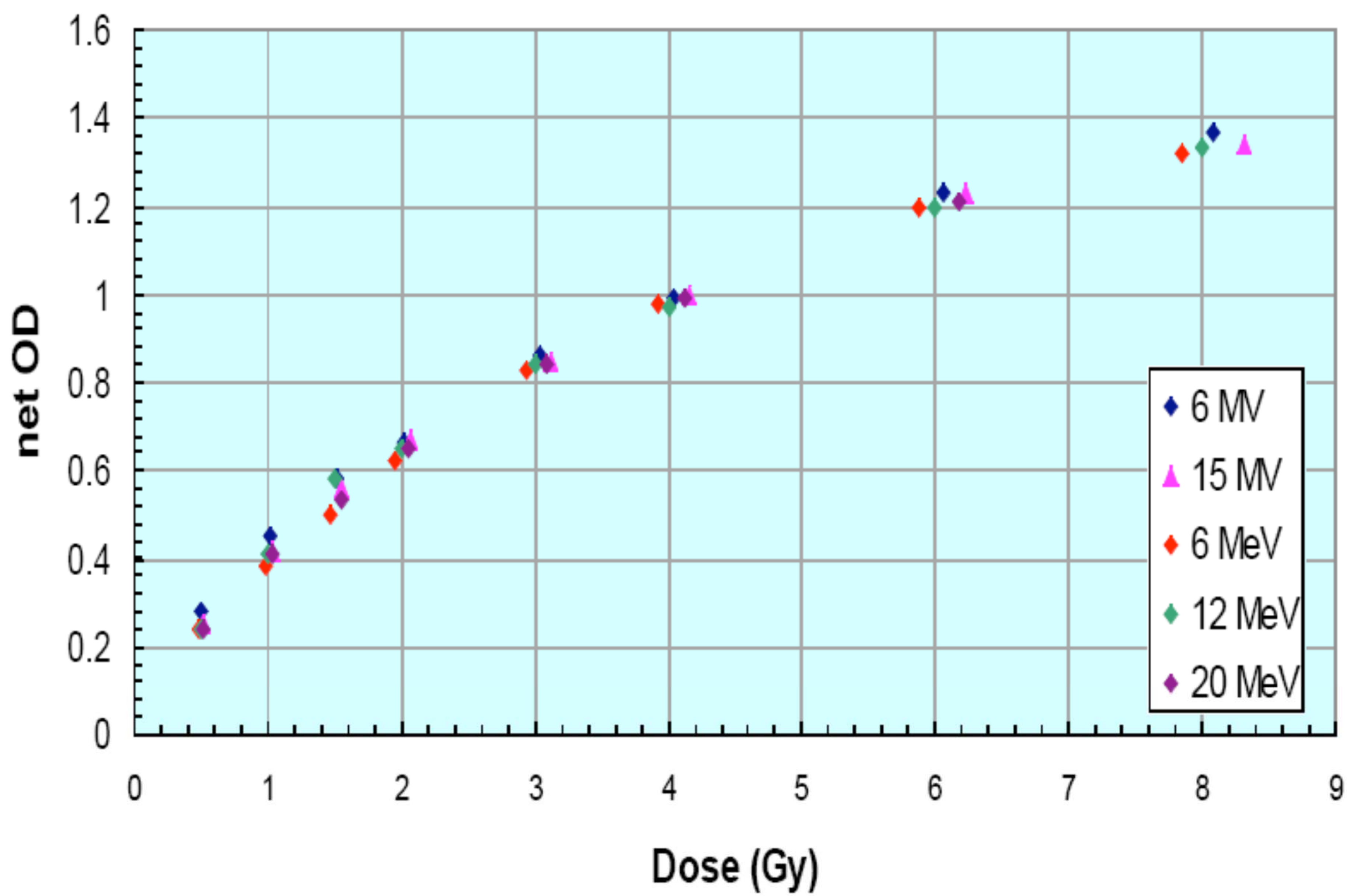
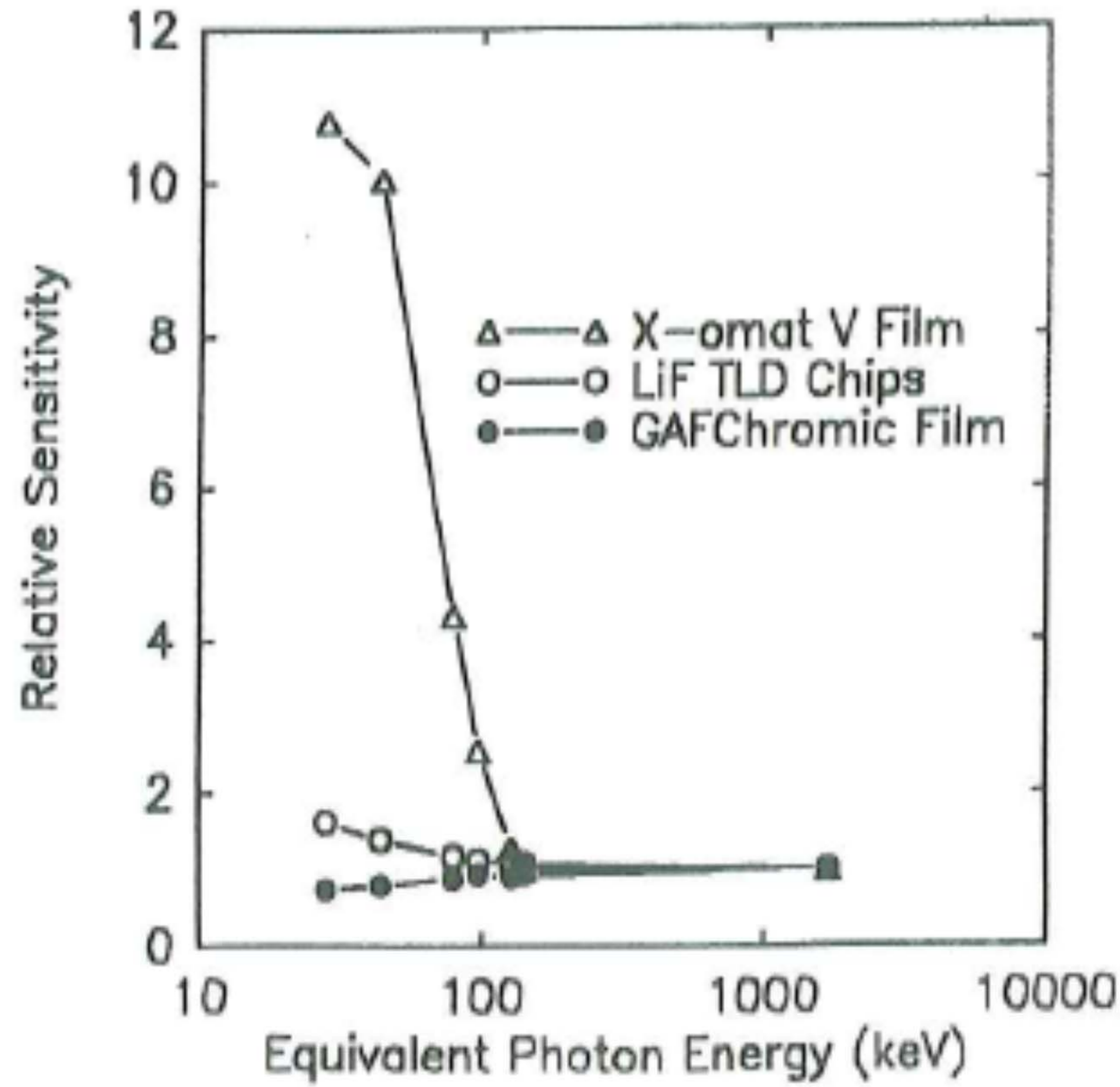


Figure 6: Sensitometric response of GAFCHROMIC® EBT in the visual and red color bands



net OD
1.6
1.4
1.2
1
0.8
0.6
0.4
0.2
0
0
1
2
3
4
5
6
7
8
9
Dose (Gy)
6 MV
15 MV
6 MeV
12 MeV
20 MeV

ISP



12
10
8
6
4
2
0
10
100
1000
10000
Relative Sensitivity
Equivalent Photon Energy (keV)
X-omat V Film
LiF TLD Chips
GAFChromic Film

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

Characteristic	Diode	MOSFET	TLD	OSLD	Film	
					Radiographic	Radiochromic
Dose	+	+	0	0	+	+
Accumulated dose	++	++	+	++	++	++
Dose rate	+	+	0	0	0	0
Energy	+[2]	+[2]	+	+	++	+
SSD	+	++	0	0	0	0
Field size	++	+	0	0	0	0
Linearity	0	0	+	+	+	+
Reproducibility	0	+	+	+	+	+
Orientation	+	+	0	0	0	0
Temperature	+	+	0	0	0	+
Readout delay	0	0	++	+	+	+
Invasive	+	+	0	0	+	+
Correction factors	++	++	+	+	+	+
Dose uncertainty (Estimated 1SD)[3]	± 3%	± 5%	± 2%	± 2%	± 3%[4]	± 3%
Main advantages	Reproducibility, Immediate readout	Immediate reading, minor fading	No cables, reused after annealing, few corrections	No cables, readout 10 min post irradiation, reused after optical bleaching	2 D dose, resolution, reread, permanent record, various shapes	2 D dose, resolution, reread, permanent record, various shapes, light insensitive
Main disadvantages	Cumbersome calibration, many corrections, cable	Limited lifetime, high cost	Labor intensive, TLD equipment	Short lifetime, dependence on accumulated dose, OSLD equipment	Light sensitive, processing equipment and maintenance, scanning equipment	Cost, scanning equipment, strict readout protocol

In-Vivo Clinical Applications

- entrance and exit doses
- skin dose
- peripheral dose
- tumor dose
- IMRT
- IGRT
- TBI & TSET
- IORT

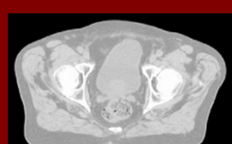
EPID dosimetry using a back-projection algorithm

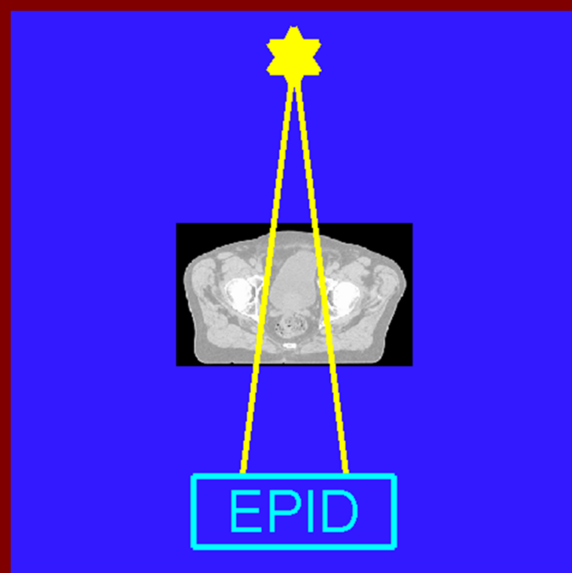
1) calculate plan

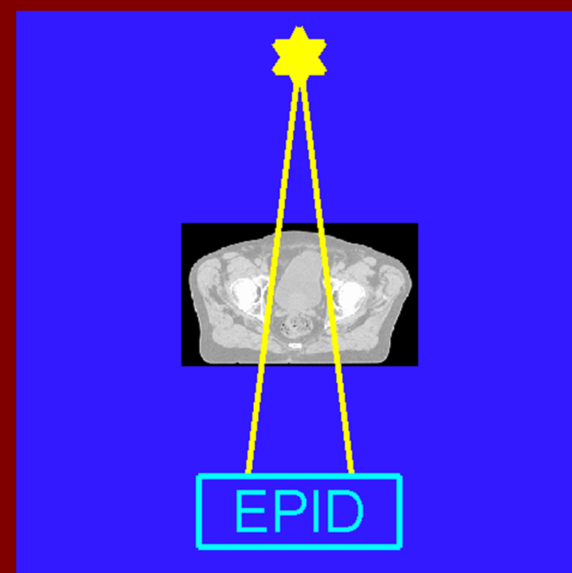
2) measure EPID dose

3) reconstruct dose in many planes for all gantry angles

patient (CT)





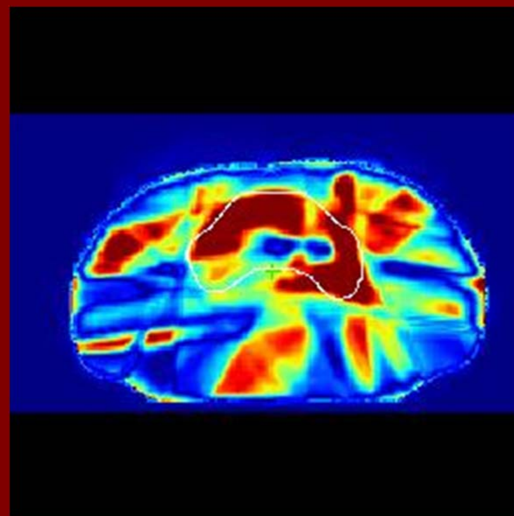


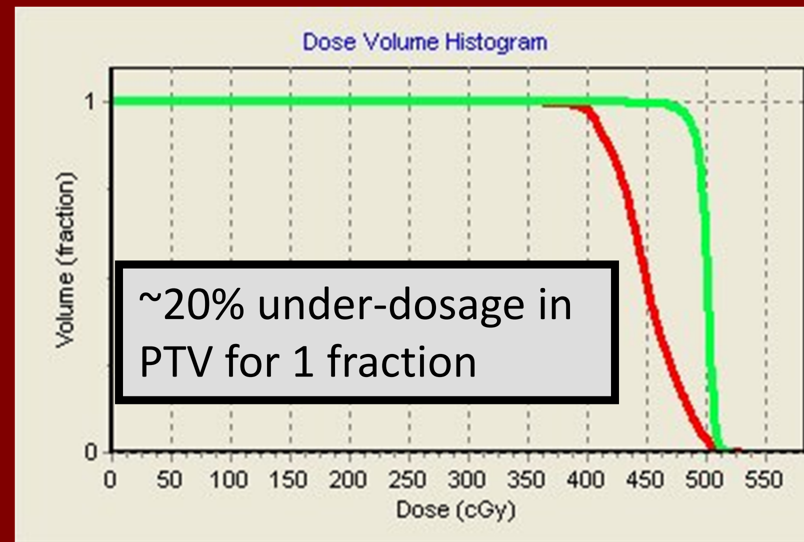
4) compare plan and reconstructed patient dose

Mihnheer et al.

Rectum IMRT: plan transfer error

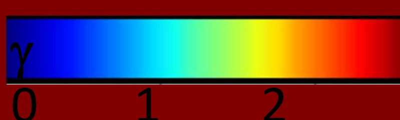
fraction 1

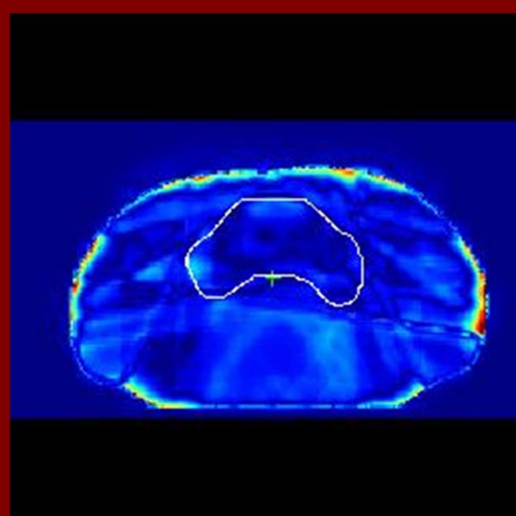


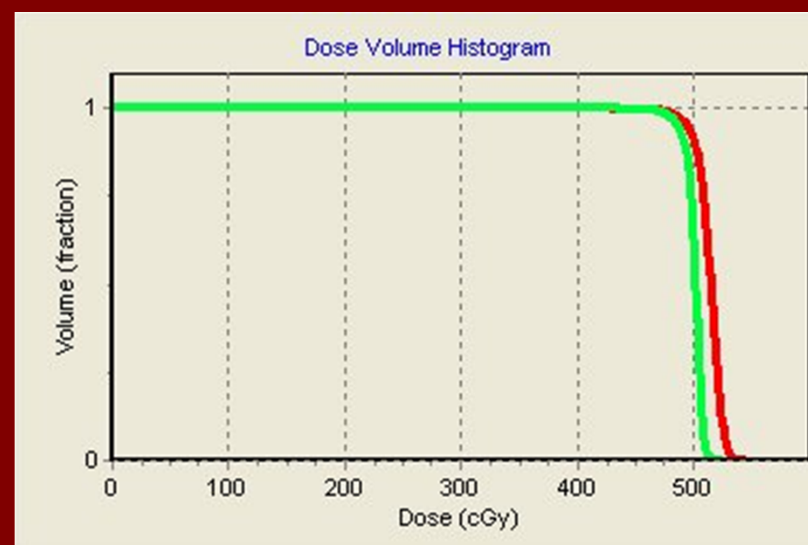


fraction 2 (plan sent again)

EPID vs. Plan





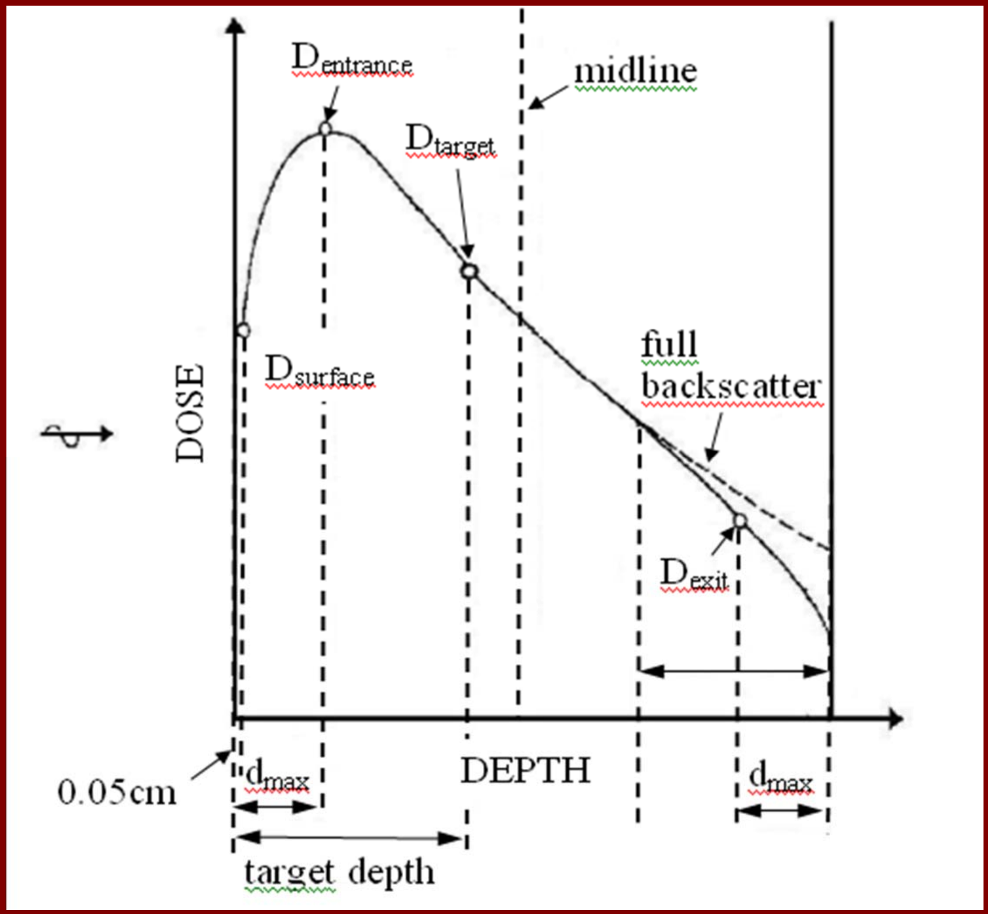


Mihnheer et al.

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

$$D_{midplane} = (D_{ent} + D_{ex})/2$$

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

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

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 atttenuation
scatter

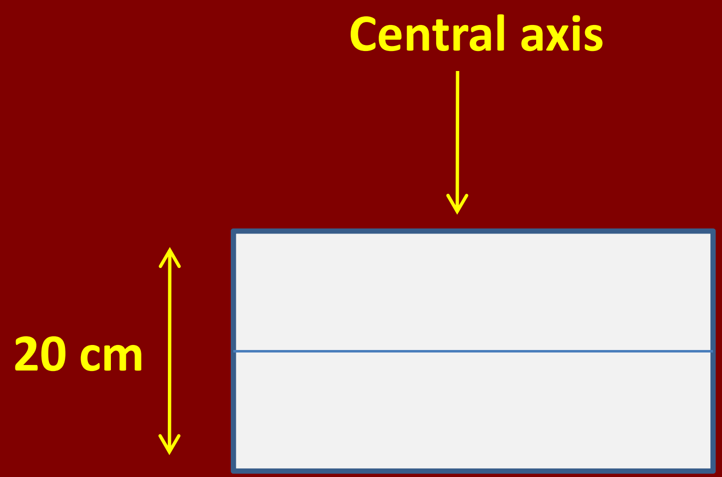


Table 1. Ratio of calculated and measured isocenter dose in a 10 × 10-cm^2 field for a homogeneous phantom with a thickness of 20 cm

Method	4 MV	8 MV	18 MV
Arithmetic mean	1.112	1.041	1.014
Geometric mean	0.948	0.969	0.975
Rizzotti	1.005	0.995	1.004
Transmission dose	1.007	0.997	0.998

The estimated uncertainty in the ratios is 1.0% (1 SD) or better.

Bollard et al

I. J. Radiation Oncology ● Biology ● Physics Volume 41, Number 2, 1998

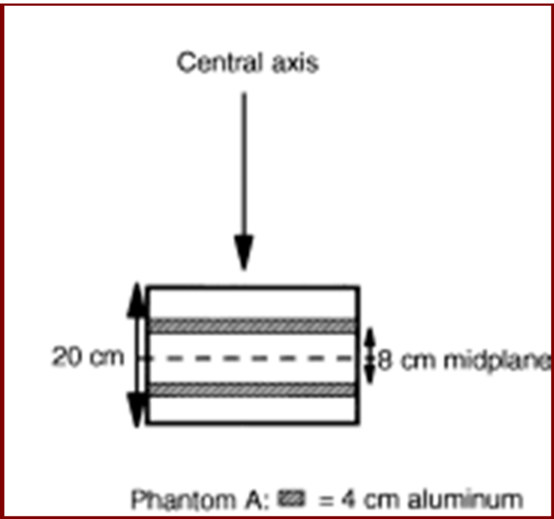


Table 2. Ratio of calculated and measured isocenter dose in a $10 \times 10\text{-cm}^2$ field for inhomogeneous phantom A, which simulates the pelvic region

Method	4 MV	8 MV	18 MV
Arithmetic mean	1.166	1.083	1.122
Geometric mean	0.940	0.980	0.978
Rizzotti	0.999	0.998	1.002
Transmission dose	1.024	1.012	1.011

The estimated uncertainty in the ratios is 1.0% (1 SD) or better.

Bollard et al.

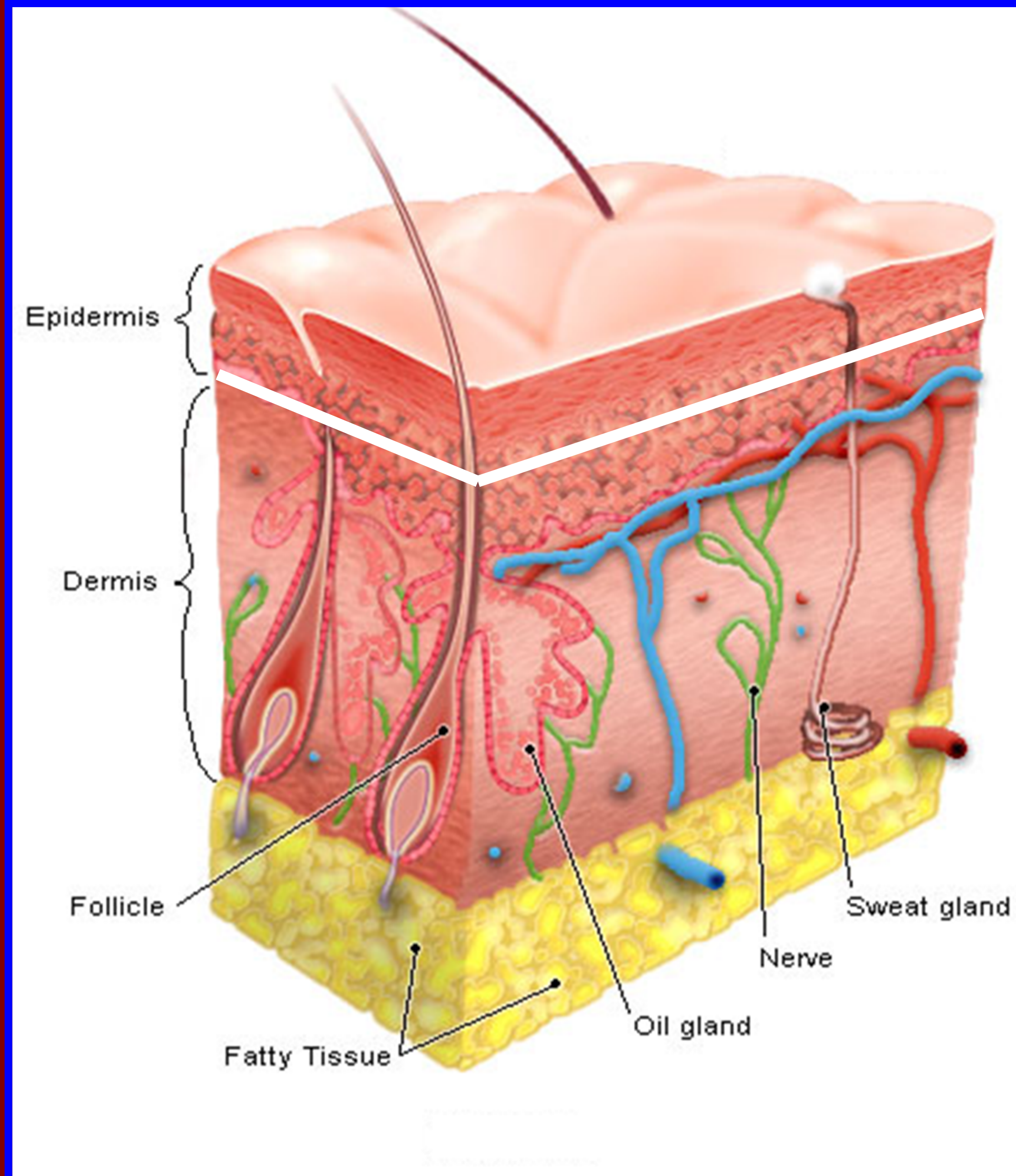
I. J. Radiation Oncology ● Biology ● Physics Volume 41, Number 2, 1998

Where the Skin Dose should be reported?

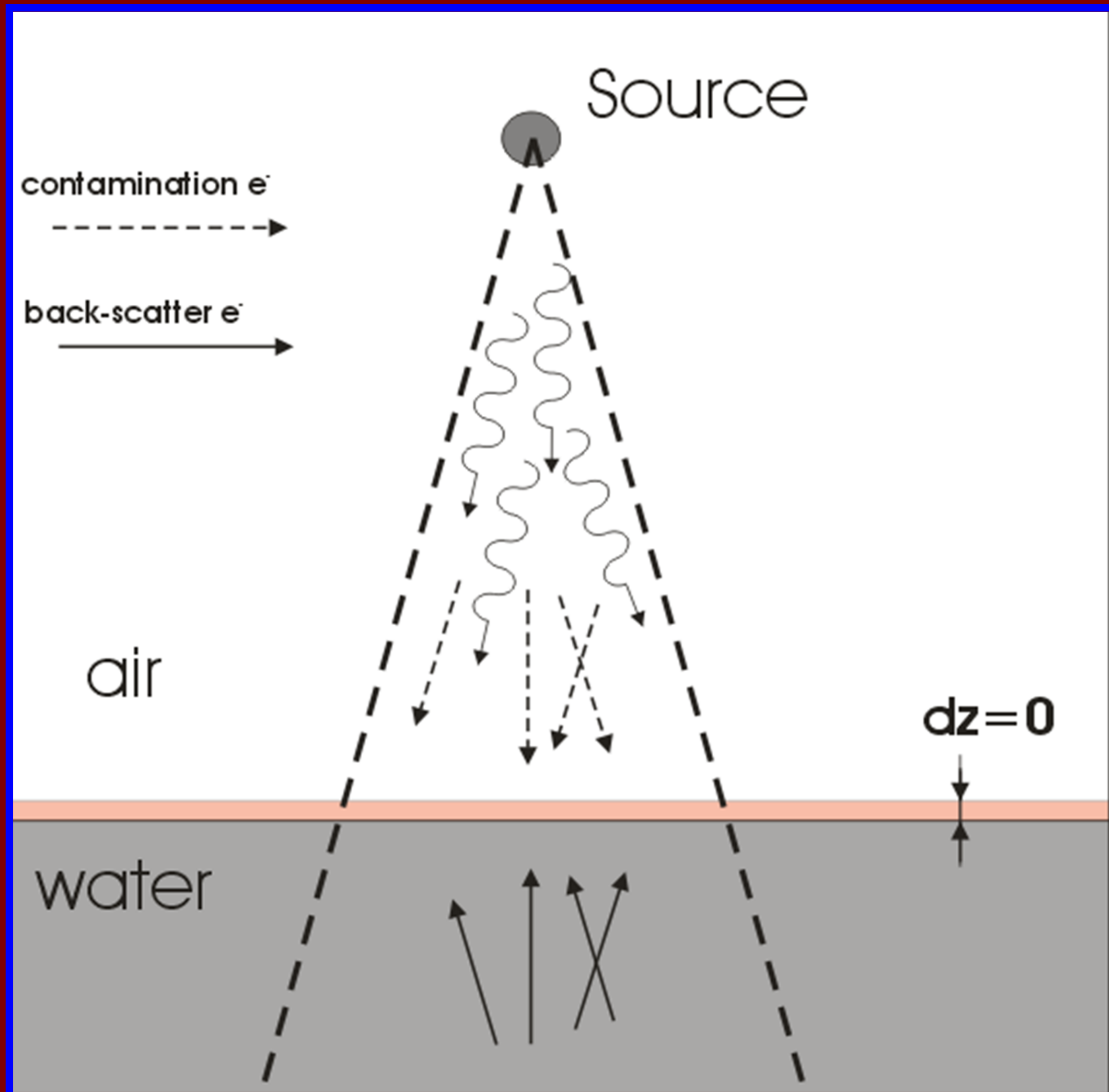
- ICRP 23 =>
- Epidermis:
  - Eyelid – 0.05 mm
  - Sole of foot – 1.5 mm

- ICRU 39 and
- ICRP 60 =>

  70 µm



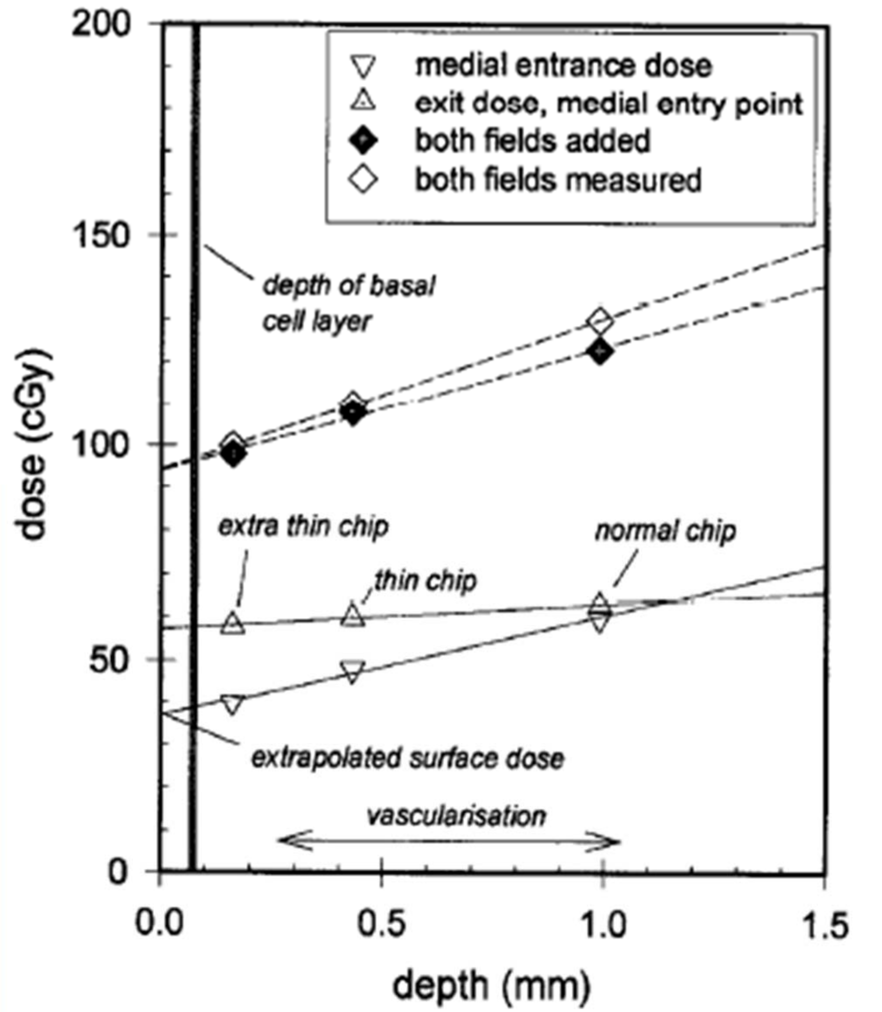
What is the Surface Dose?



Surface Dose = Skin Dose ?

Dosimeter	Depth scaled by density (mm)	Correction
Attix chamber	0.048	1.062
Extrapolation chamber	0.069	1.000
HS GAFCHROMIC® film	0.153	0.850
EBT GAFCHROMIC® film	0.153	0.854
XR-T GAFCHROMIC® film	0.157	0.837
TLD-100 (0.15 mm)	0.185	0.810
TLD-100 (0.4 mm)	0.496	0.586

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.



T. Kron et al. / Radiotherapy and Oncology 41 (1996) 119–123

Quach et al.: Measurement of radiotherapy x-ray skin dose
Med. Phys. 27 .7., July 2000

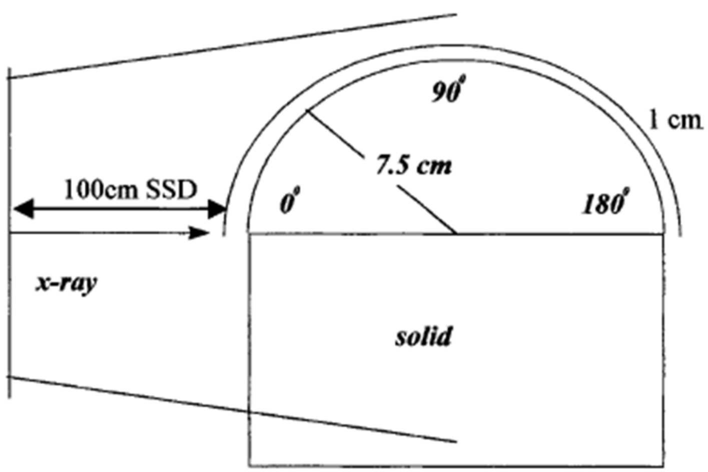


FIG. 1. The hemicylindrical chest wall phantom used to simulate a tangential breast treatment. A 20×20 cm^2 x-ray beam with 2.5 cm overshoot was set incident at 100 cm SSD, for some of the radiochromic film readings 1 cm of bolus was added.

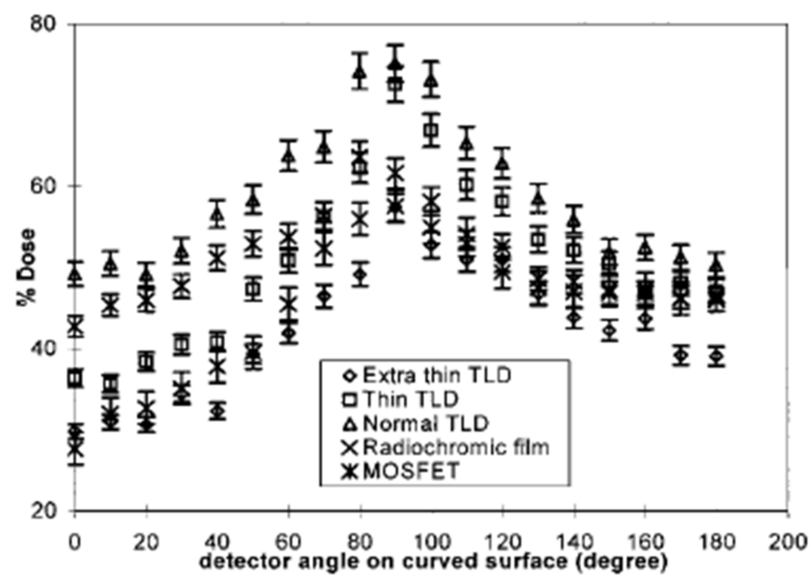


FIG. 2. Percentage near surface dose obtained using various detectors including radiochromic (Gafchromic MD 55-2) film, TLDs (extra thin, thin and normal), and a MOSFET detector. Normalized to incident dose at d_{max} for each detector.

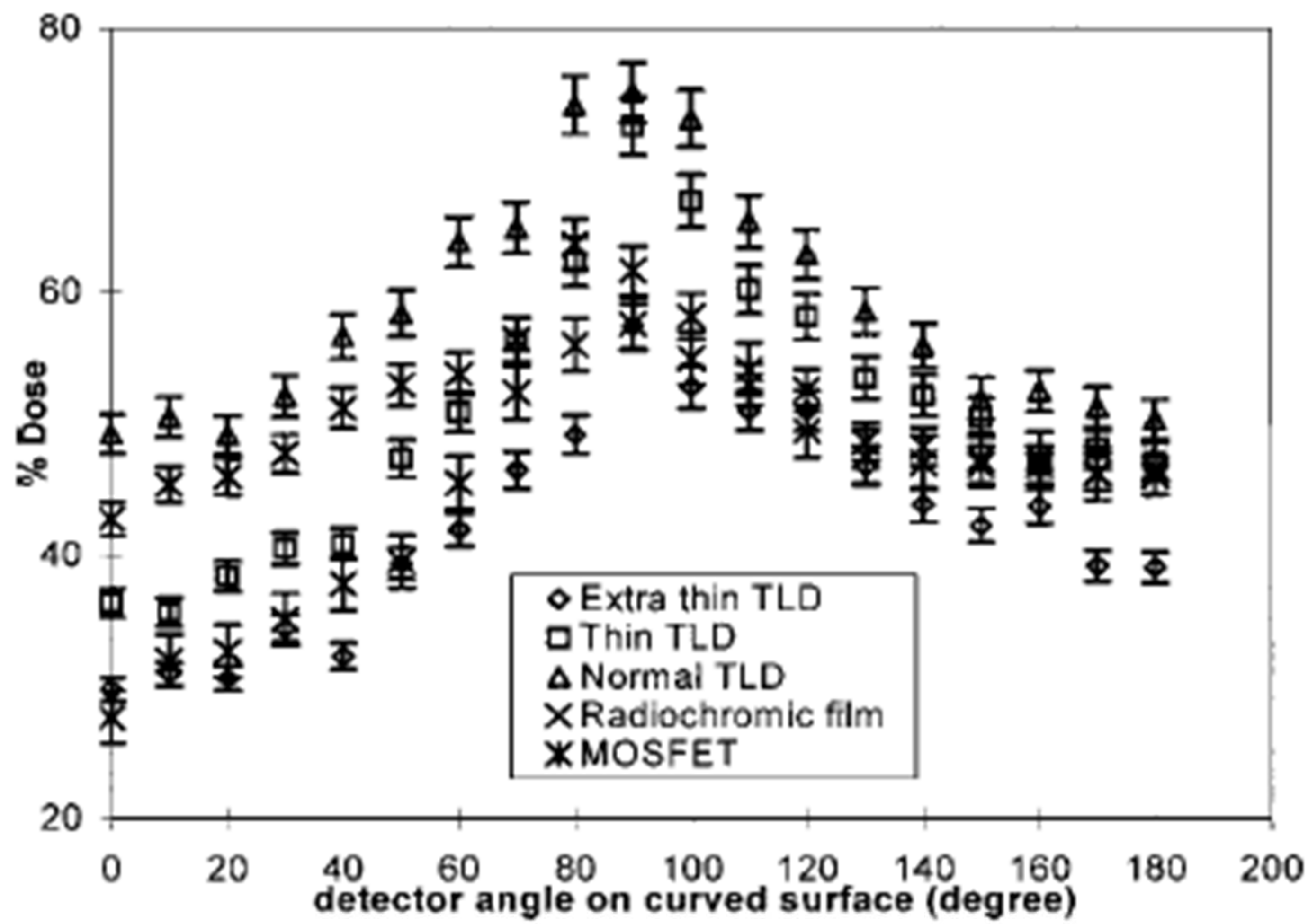


FIG. 2. Percentage near surface dose obtained using various detectors including radiochromic (Gafchromic MD 55-2) film, TLDs (extra thin, thin and normal), and a MOSFET detector. Normalized to incident dose at d_{max} for each detector.

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

Int. J. Radiation Oncology Biol. Phys., Vol. 61, No. 5, pp. 1603–1606, 2005
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0360-3016/05/$–see front matter

doi:10.1016/j.ijrobp.2004.12.034

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.

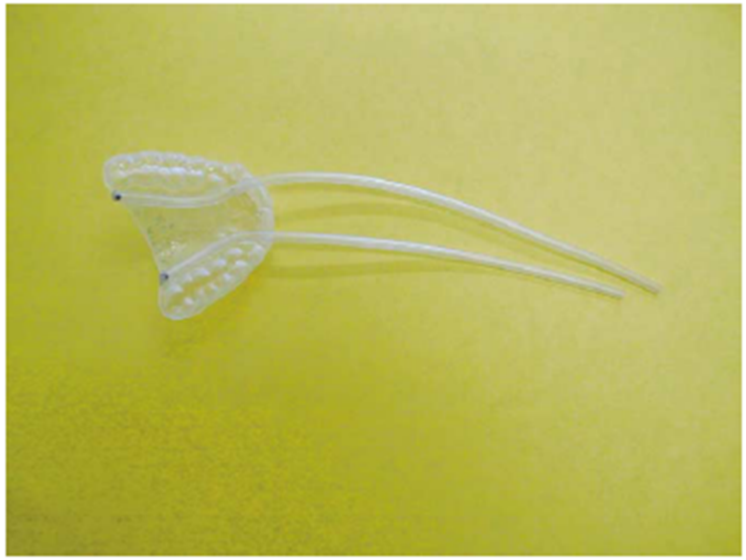

Fig. 1. The molded mouth plate with the two catheters.

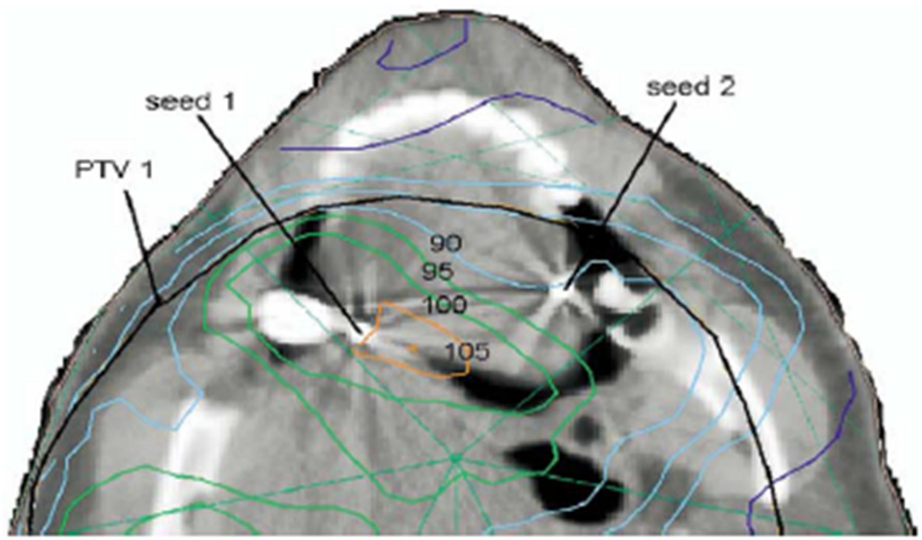

Fig. 2. Image of seeds on a computed tomography slice.

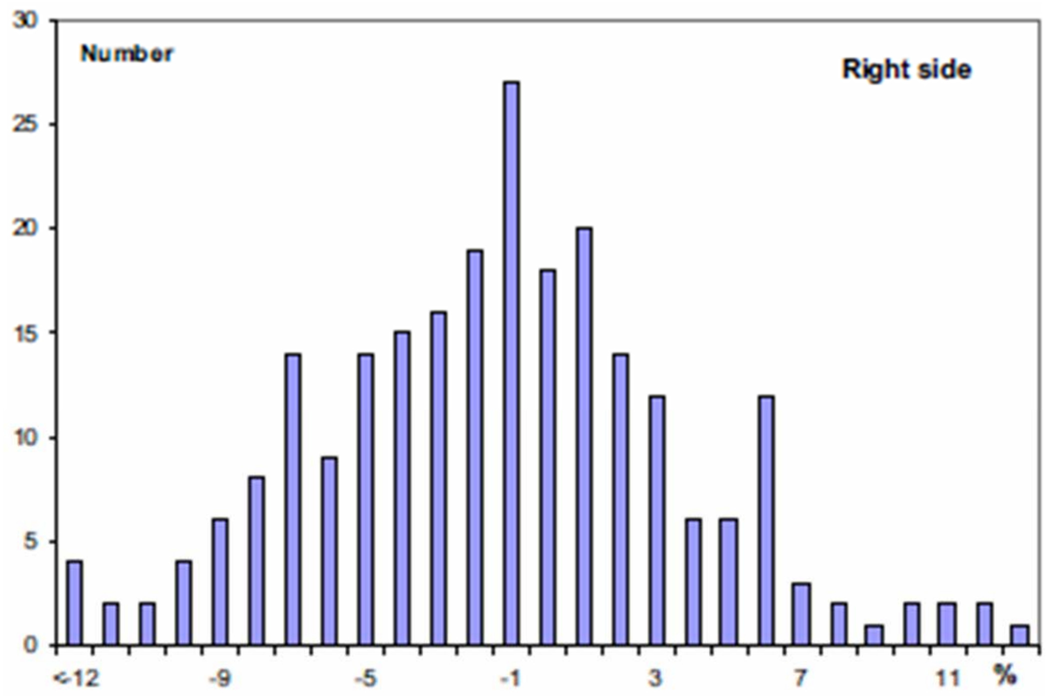

Fig. 3. Number of measurements with discrepancies (in percentages) between calculated and measured values for the right side.

Agreement of measurements with calculations within 5% despite uncertainties due to interfraction patient positions, intrafraction patient movements, and interfraction MOSFET positions.

In vivo measurements with MOSFET ● S. Marcié *et al.*
I. J. Radiation Oncology ● Biology ● Physics Volume 61, Number 5, 2005

JOURNAL OF APPLIED CLINICAL MEDICAL PHYSICS, VOLUME 7, NUMBER 4, FALL 2006
22 © 2006 Am. Coll. Med. Phys. 22

In vivo prostate IMRT dosimetry with MOSFET detectors
using brass buildup caps

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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 (*Dmax) from the prostate* IMRT fields.** A general formalism with various correction factors taken into account
to predict *Dmax 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 ±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%.

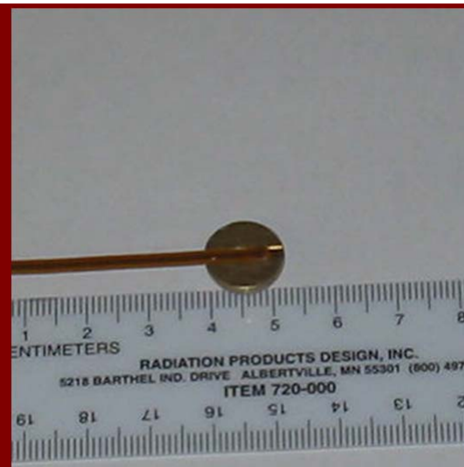


Table 2. *In vivo* measurements using MOSFETs compared to corrected target doses from the TPS (CMS XiO). All plans delivered with beam energy 10 MV. (CR = 1.038)

Patient	Field #	D_{max} Dose TPS (cGy)	IMRT CF	Total CF	Expected MOSFET dose (cGy)	Measured MOSFET dose (cGy)	% Diff
	2	74.6	0.996	1.036	77.3	78.5	1.6%
#1	3	49.0	0.992	1.049	51.4	52.6	2.2%
	4	26.7	0.975	1.020	27.3	28.7	5.1%
	5	43.7	0.977	1.024	44.7	46.1	3.1%
	6	84.2	1.012	1.053	88.6	91.1	2.8%
Total		**278.2**			**289.3**	**296.9**	**2.6%**
	2	61.1	1.018	1.071	65.4	65.2	–0.4%
#2	3	44.7	0.996	1.009	45.1	44.0	–2.4%
	4	34.3	1.026	1.044	35.8	35.8	–0.1%
	5	39.1	1.004	1.017	39.8	39.3	–1.3%
	6	62.8	0.991	1.069	67.1	70.0	4.3%
Total		**242.0**			**253.3**	**254.3**	**0.4%**

In vivo dose verification of IMRT treated head and neck cancer patients

PER E. ENGSTRO¨ M, PIA HARALDSSON, TORSTEN LANDBERG,
HANNE SAND HANSEN, SVEND AAGE ENGELHOLM & HA°
KAN NYSTRO¨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

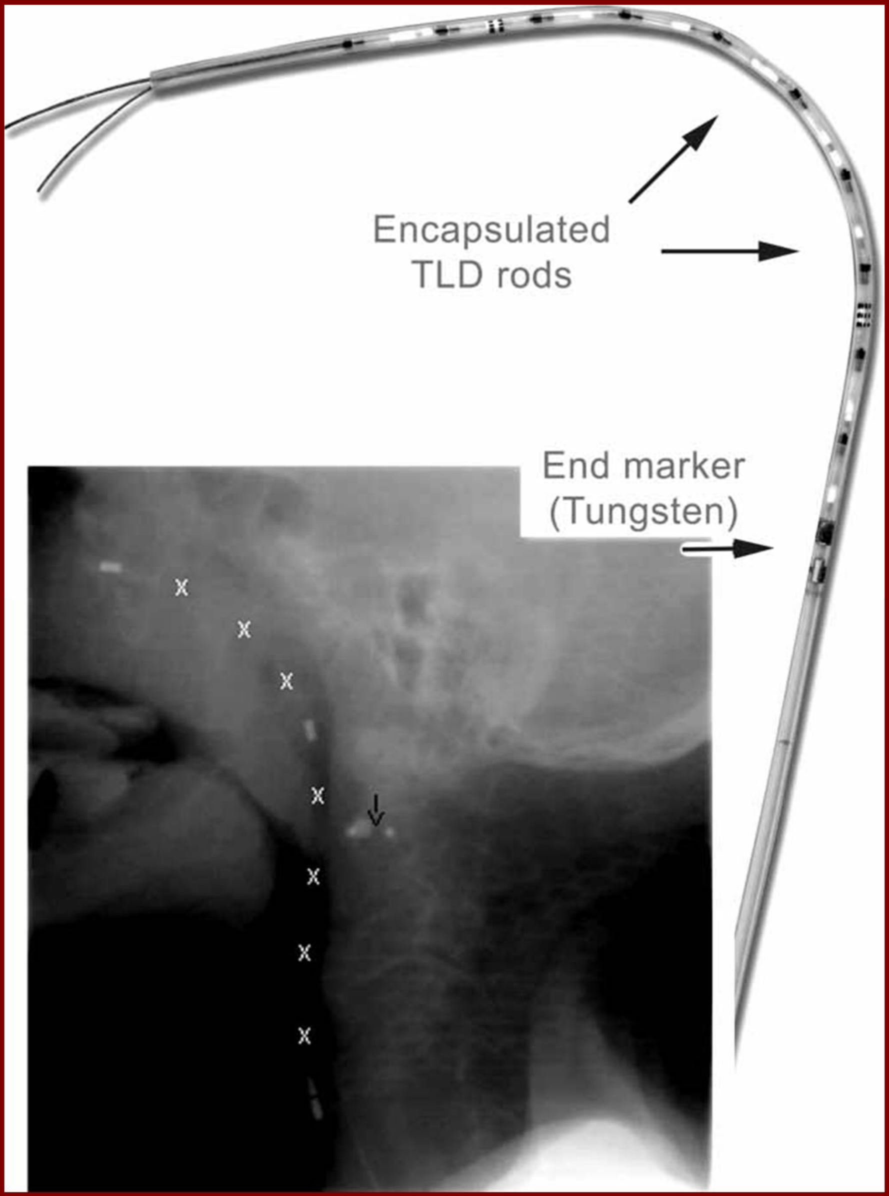


Figure 1. Photo of the tube containing ten encapsulated TLD rods. The tube extends several centimetres beyond the tungsten ''end markers'' only for the purpose of the patients' comfort. The total length of the tube is approximately 30 cm but depends on anatomy and diagnose. Inserted picture (lower left) 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

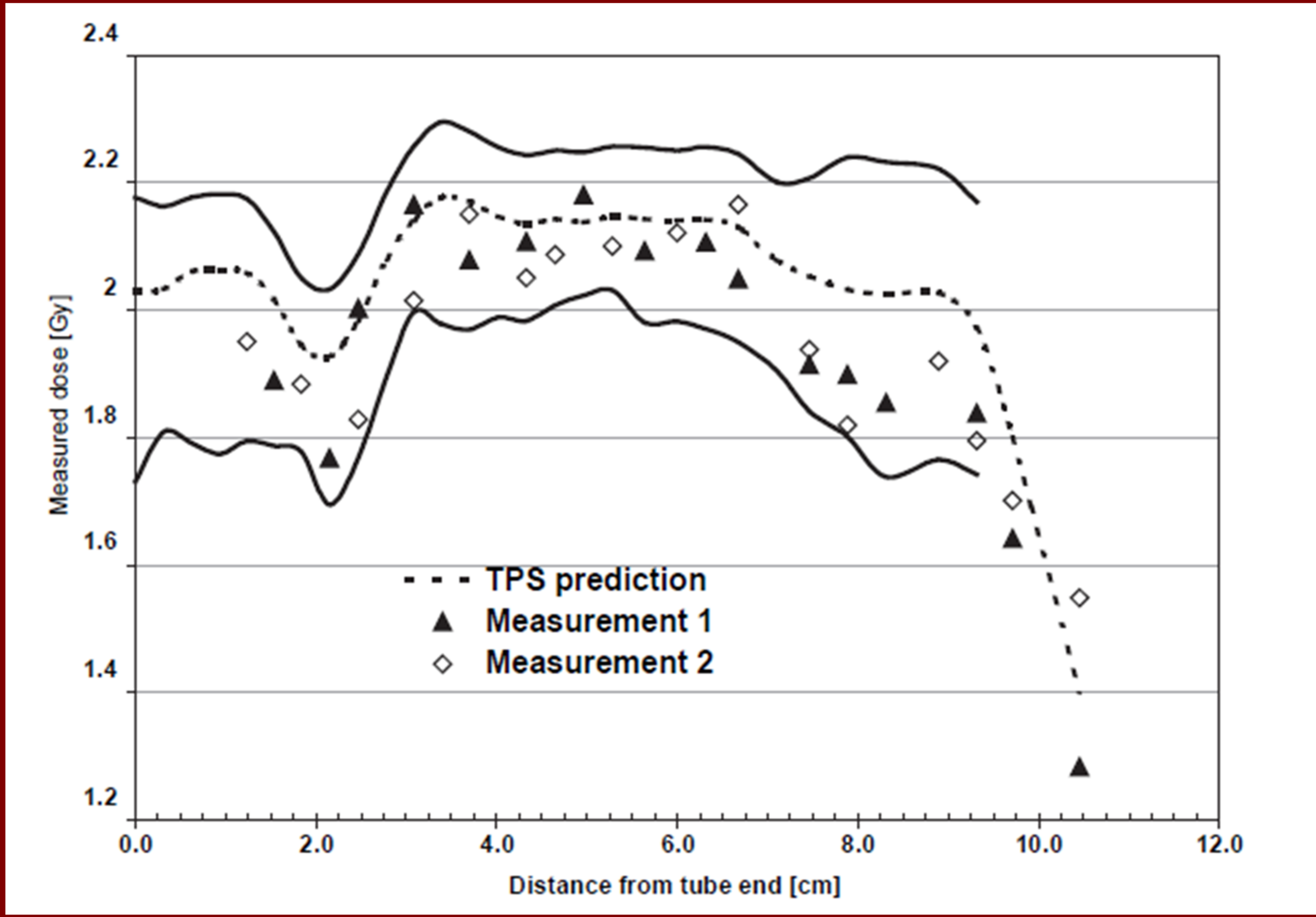


Figure 3. TLD readings of two measurements performed on a rhinopharynx patient with one-week interval. The solid lines represent upper and lower tolerance levels (dosimetric + geometric).

P. Engstrom et al., ACTA Oncologica, 44 (2005) 572

Table I. Comparison between OSLD and TLD dose measurements for TBI patients

Location	Patient #1	Patient #2	D(cal)/D(meas) Pat 1 : Pat 2
	OSLD/TLD	OSLD/TLD	
Head	0.99	0.99	1.04 : 0.99
Umbilicus	0.97	0.97	0.99 : 1.02
Feet	0.98	0.97	1.03 : 1.05
< (OSLD/TLD)> = 0.98 ± 0.01			

Determination of 2 Fields to 12 Fields factor for TSEI using TLDs and OSLDs

Method: 3 TLDs and 3 OSLDs located at 60^0 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_{eff} = 3.4 MeV at phantom surface

Deliver 290 MUs for 12 fields → 270 ± 17^0 for 6 phantom orientations of $0^0 – 60^0 – 120^0 – 180^0 – 240^0 - 300^0$

	2 fld Gy	12 fld Gy	12 fld / 2 fld
OSLD	0.360	1.09	3.03 ± 0.11
TLD	0.378	1.14	3.02 ± 0.11

Table II. Comparison between OSLD and TLD dose measurements for TSEI patients. Values in parentheses are D(cal) / D(measured)

Location	Patient #1	Patient #2
	OSLD / TLD	OSLD / TLD
Forehead	1.02 (1.13)	1.01 (1.18)
Chest	0.97 (1.10)	0.94 (1.12)
Anterior	0.97 (1.11)	1.01 (1.06)
Posterior	1.04 (1.09)	1.03 (1.09)
Right lateral	1.02 (1.01)	0.97 (0.96)
Left lateral	1.02 (1.05)	0.97 (0.92)
Mid thigh	0.97 (1.11)	1.02 (1.04)
Mid shin	0.97 (1.03)	0.98 (1.05)
$<(\text{OSLD/TLD})> = 0.99 \pm 0.03$		

Out-of-Field Dose

Requirements:

- **Sensitivity**
- **Energy Dependence**
- **Size**

↓

TLDs or OSLDs

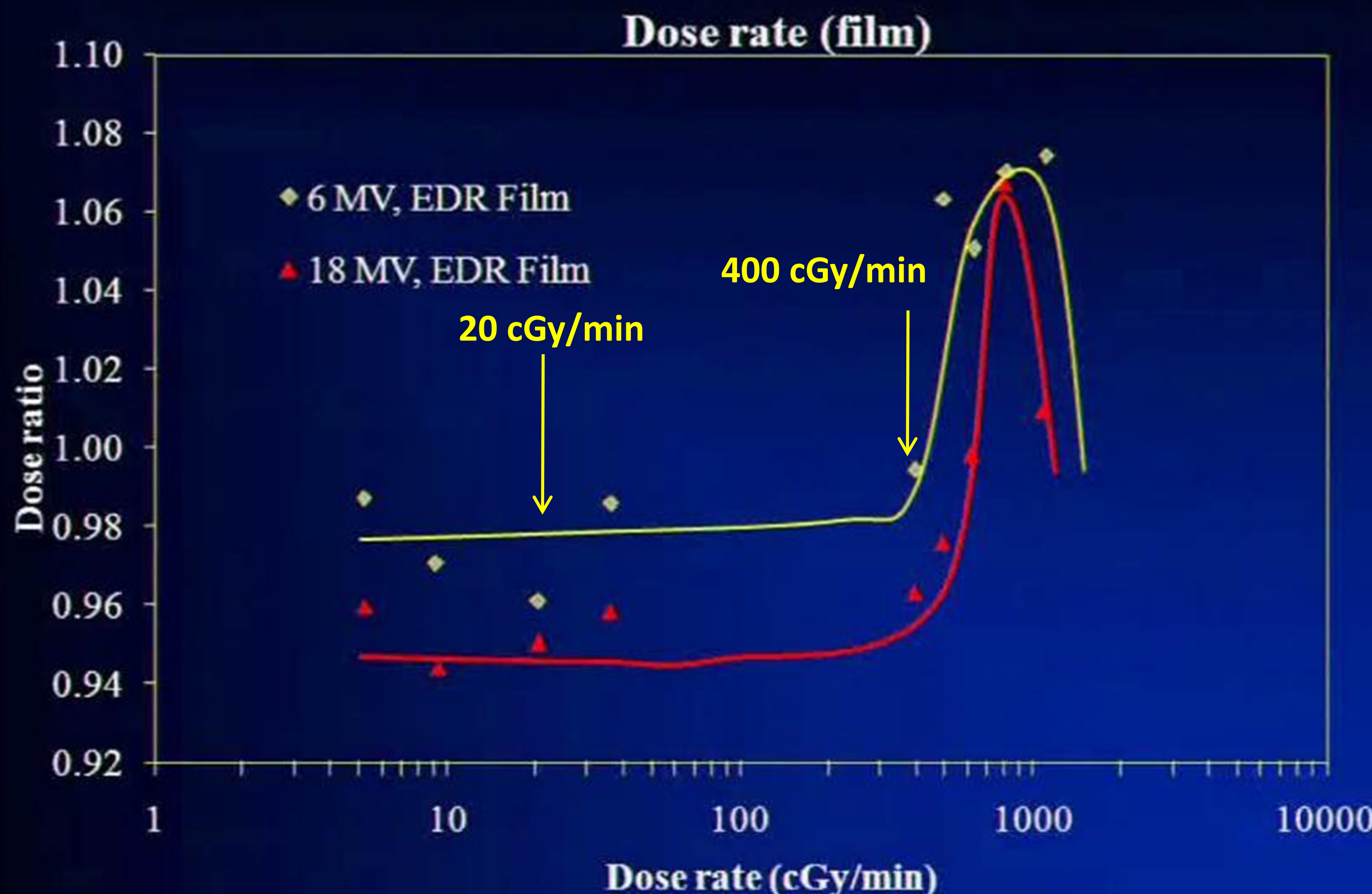
CONCLUSIONS

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 ???

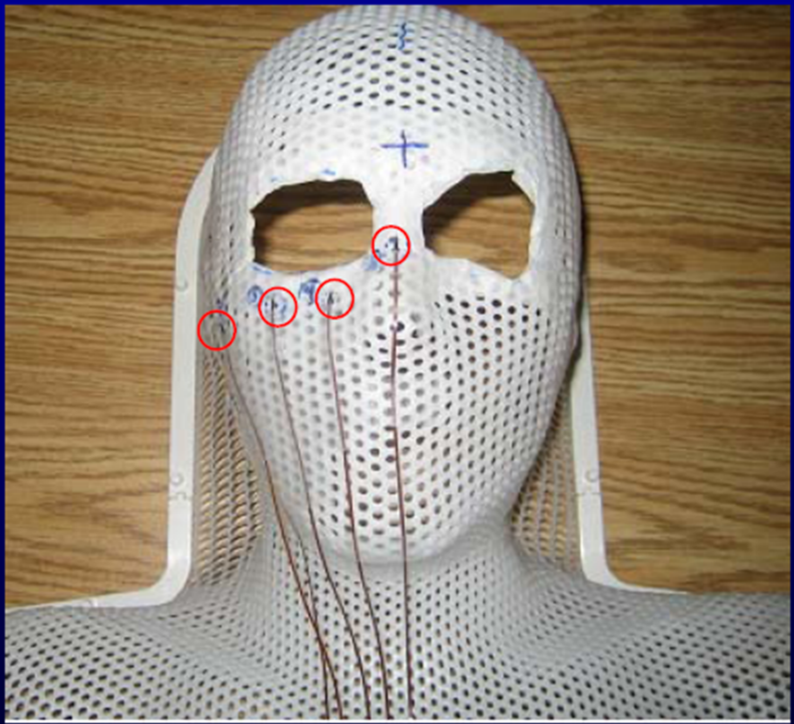


Srivastava and Das *Med Phys* 33:2089 , 2006

INDIANA UNIVERSITY MPRI

IMRT in vivo dosimetry

Marcie et al, Int. J. Rad. Onc. Biol. Phys., 61, 1603-6, 2005



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