



Standard Guide for Selecting Dosimetry Systems for Application in Pulsed X-Ray Sources¹

This standard is issued under the fixed designation E 1894; the number immediately following the designation indicates the year of original adoption or, in the case of revision, the year of last revision. A number in parentheses indicates the year of last reapproval. A superscript epsilon (ε) indicates an editorial change since the last revision or reapproval.

1. Scope

1.1 This guide provides assistance in selecting and using dosimetry systems in flash X-ray experiments. Both dose and dose-rate techniques are described.

1.2 Operating characteristics of flash x-ray sources are given, with emphasis on the spectrum of the photon output.

1.3 Assistance is provided to relate the measured dose to the response of a device under test (DUT). The device is assumed to be a semiconductor electronic part or system.

2. Contents

2.1 Section 1: Scope of guide.

2.2 Section 2: Outline.

2.3 Section 3: Related ASTM and ICRU documents.

2.4 Section 4: Definition of terms.

2.5 Section 5: Significance and use of this document for the selection of dosimetry systems for use in pulsed x-ray sources.

2.6 Section 6: Description of large flash x-ray sources and their characteristics.

2.7 Section 7: Measurement principles with an emphasis on obtaining absorbed dose measurements for different spectral conditions in the dosimeter, the DUT, and the relationship between them.

2.8 Section 8: The primary information in this guide. The experimenter will find details on each dosimetry system. Listed are details such as: 1) how the dosimeters work, i.e., physical principles, 2) typical applications or instrumentation configurations, 3) advantages, 4) limitations, 5) sensitivity 6) procedures for calibration and proper use and finally reproducibility and accuracy.

2.9 Section 9: Suggested documentation requirements.

2.10 Section 10: Description of how the experimenter determines uncertainty in the dosimetry measurements.

2.11 Section 11: References.

3. Referenced Documents

3.1 *ASTM Standards*:²

E 170 Terminology Relating to Radiation Measurements and Dosimetry

E 665 Practice for Determining Absorbed Dose Versus Depth in Materials Exposed to the X-ray Output of Flash X-ray Machines³

E 666 Practice for Calculating Absorbed Dose From Gamma or X Radiation

E 668 Practice for Application of Thermoluminescence-Dosimetry (TLD) Systems for Determining Absorbed Dose in Radiation-Hardness Testing of Electronic Devices

E 1249 Practice for Minimizing Dosimetry Errors in Radiation Hardness Testing of Silicon Electronic Devices Using Co-60 Sources

E 1261 Guide for Selection and Calibration of Dosimetry Systems for Radiation Processing³

E 1275 Practice for Use of a Radiochromic Film Dosimetry System³

E 1310 Practice for Use of a Radiochromic Optical Waveguide Dosimetry System³

3.2 *International Commission on Radiation Units (ICRU) and Measurements Reports*:⁴

ICRU Report 14 Radiation Dosimetry: X rays and Gamma Rays with Maximum Photon Energies Between 0.6 and 50 MeV

ICRU Report 17 Radiation Dosimetry: X rays Generated at Potentials of 5 to 150 kV

¹ This practice is under the jurisdiction of ASTM Committee E10 on Nuclear Technology and Applications and is the direct responsibility of Subcommittee E10.07 on Radiation Dosimetry for Radiation Effects on Materials and Devices.

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² For referenced ASTM standards, visit the ASTM website, www.astm.org, or contact ASTM Customer Service at service@astm.org. For *Annual Book of ASTM Standards* volume information, refer to the standard's Document Summary page on the ASTM website.

³ Withdrawn. The last approved version of this historical standard is referenced on www.astm.org.

⁴ Available from the International Commission on Radiation Units and Measurements, 7910 Woodmont Ave., Suite 800, Bethesda, MD 20814, U.S.A.

ICRU Report 33 Radiation Quantities and Units
ICRU Report 34 The Dosimetry of Pulsed Radiation

4. Terminology

4.1 *absorbed dose*—quotient of $d\bar{e}/dm$, where $d\bar{e}$ is the mean energy imparted by ionizing radiation to matter of mass dm :

$$D = \frac{d\bar{e}}{dm} \quad (1)$$

See ICRU Report 33. The special name for the unit for absorbed dose is the gray (Gy).

$$1 \text{ Gy} = 1\text{J/kg} \quad (2)$$

Formerly, the special unit for absorbed dose was the rad, where 1 rad = 100 erg/g.

$$1 \text{ rad} = 0.01 \text{ Gy} \quad (3)$$

Because the magnitude of the absorbed dose is material dependent, it is important to include the material composition for which the dose is being reported, e.g., 15.3 Gy(LiF).

4.2 *absorbed dose enhancement*—increase (or decrease) in the absorbed dose (as compared to the equilibrium absorbed dose) at a point in a material of interest. This can be expected to occur near an interface with a material of higher or lower atomic number.

4.3 *converter*—a target for electron beams, generally of a high atomic number material, in which bremsstrahlung X rays are produced by radiative energy losses of the incident electrons.

4.4 *dosimeter*—a device that, when irradiated, exhibits a quantifiable change in some property of the device which can be related to absorbed dose in a given material using appropriate analytical instrumentation and techniques.

4.5 *dosimetry system*—A system used for determining absorbed dose, consisting of dosimeters, measurement instruments, and their associated reference standards, and procedures for the system's use.

4.6 *DUT*—device under test. This is the electronic component or system tested to determine its performance during or after irradiation.

4.7 *endpoint energy*—endpoint energy refers to the peak energy of the electron beam, usually in MeV, generated in a flash X-ray source and is numerically equal to the maximum voltage in MV. The word endpoint refers to the highest photon energy of the bremsstrahlung spectra, and this endpoint is equal to the maximum or peak in the electron energy. For example, if the most energetic electron that strikes the converter is 10 MeV, this electron produces a range of bremsstrahlung photon energies but the maximum energy of any photon is equal to 10 MeV, the endpoint energy. Most photons have energies one-tenth to one-third of the maximum electron energy for typical flash X-ray sources in the 10 MV to 1 MV endpoint voltage region, respectively.

4.8 *endpoint voltage*—Endpoint voltage refers to the peak voltage across a bremsstrahlung diode in a flash X-ray source. For example, a 10-MV flash X-ray source is designed to reach a peak voltage of 10-MV across the anode-cathode gap which generates the electron beam for striking a converter to produce bremsstrahlung.

4.9 *equilibrium absorbed dose*—absorbed dose at some incremental volume within the material in which the condition of electron equilibrium (the energies, number, and direction of charged particles induced by the radiation are constant throughout the volume) exists. For lower energies where bremsstrahlung production is negligible the equilibrium absorbed dose is equal to the kerma.

NOTE 1—For practical purposes, assuming the spatial gradient in the X-ray field is small over the range of the maximum energy secondary electrons generated by the incident photons, the equilibrium absorbed dose is the absorbed dose value that exists in a material at a distance from any interface with another material greater than this range.

5. Significance and Use

5.1 Flash X-ray facilities provide intense bremsstrahlung radiation environments, usually in a single sub-microsecond pulse, which unfortunately, often fluctuates in amplitude, shape, and spectrum from shot to shot. Therefore, appropriate dosimetry must be fielded on every exposure to characterize the environment, see ICRU Report 34. These intense bremsstrahlung sources have a variety of applications which include the following:

5.1.1 Generation of X-ray and gamma-ray environments similar to that from a nuclear weapon burst.

5.1.2 Studies of the effects of X rays and gamma rays on materials.

5.1.3 Studies of the effects of radiation on electronic devices such as transistors, diodes, and capacitors.

5.1.4 Vulnerability and survivability testing of military systems and components.

5.1.5 Computer code validation studies.

5.2 This guide is written to assist the experimenter in selecting the needed dosimetry systems (often in an experiment not all radiation parameters must be measured) for use at pulsed X-ray facilities. This guide also provides a brief summary of the information on how to use each of the dosimetry systems. Other guides (see Section 3) provide more detailed information on selected dosimetry systems in radiation environments and should be consulted after an initial decision is made on the appropriate dosimetry system to use. There are many key parameters which describe a flash X-ray source, such as dose, dose rate, spectrum, pulse width, etc., such that typically no single dosimetry system can measure all the parameters simultaneously.

6. General Characteristics of Flash X-ray Sources

6.1 *Flash X-ray Facility Considerations*—Flash X-ray sources operate like a dental X-ray source but at much higher voltages and intensities and usually in a single, very short burst, see ICRU Report 17. A high voltage is developed across an anode-cathode gap (the diode) and field emission creates a pulsed electron beam traveling from the cathode to the anode. A high atomic-number element such as tantalum is placed on the anode to maximize the production of bremsstrahlung created when the electrons strike the anode. Graphite or Aluminum is usually placed downstream of the converter to stop the electron beam completely but let the X-radiation pass through. Finally, a debris shield made of Kevlar or low-density

DIFFERENTIAL BREMSSTRAHLUNG SPECTRA

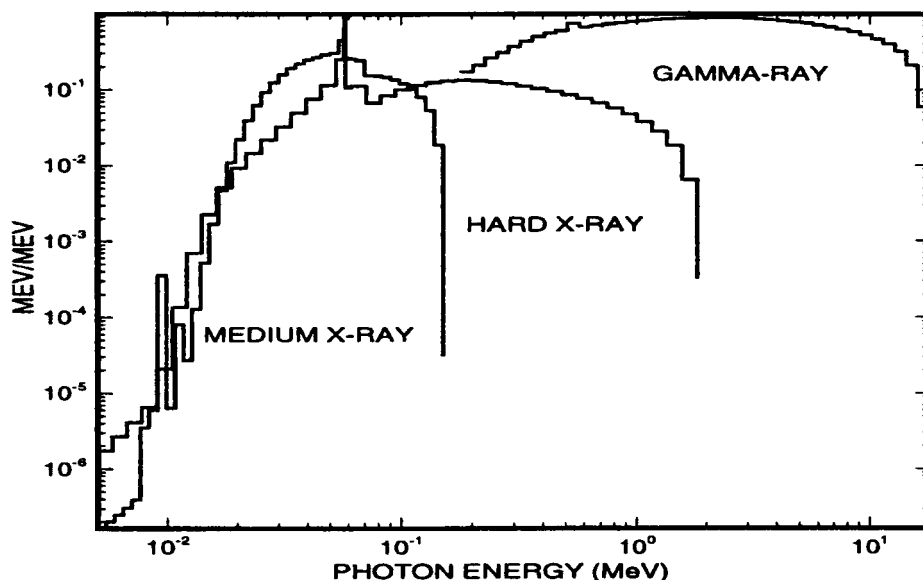


FIG. 1 Range of Available Bremsstrahlung Spectra from Flash X-ray Sources

polyethylene is sometimes necessary to stop exploding converter material from leaving the source. All of these components taken together form what is commonly called a bremsstrahlung diode.

6.2 Relationship Between Flash X-ray Diode Voltage and X-ray Energy of Bremsstrahlung—Flash X-ray sources produce bremsstrahlung by generating an intense electron beam which then strikes a high atomic number (Z) converter such as tantalum. The electron-solid interactions produce “braking” radiation or, in German, bremsstrahlung. Fig. 1 shows the typical range of photon energies produced by three different sources. If the average radiation produced is in the 20–100 keV region, the source is said to be a medium–hard X-ray simulator. If the average photon energy is in the 100–300-keV region, the term used is “hard X-ray simulator.” At the high end of the flash X-ray range are sources which produce an average photon energy of around 2 MeV. Because this photon energy is in the typical gamma-ray spectral range, the source is called a gamma-ray simulator.

6.2.1 The average energy of the bremsstrahlung spectrum, $\bar{E}_{\text{photon}}$, through an optimized converter **(1)**⁵ in the medium–hard X-ray region ($50 \text{ keV} < \bar{E}_{\text{photon}} < 500 \text{ keV}$) is given empirically by,

$$\bar{E}_{\text{photon}} \approx 5 \varepsilon^{1/2} \quad (4)$$

where $\bar{E}_{\text{photon}}$ is the average energy of the bremsstrahlung photons in keV and ε is the average energy of the electrons in the electron beam incident on the converter in keV. This equation and Fig. 1 indicate that most of the photons have energies much less than the endpoint electron energy, or in voltage units, the flash X-ray voltage.

⁵ The boldface numbers in parentheses refer to the list of references at the end of this standard.

7. Measurement Principles

7.1 Typically in flash X-ray irradiations, one is interested in some physical change in a critical region of a device under test (DUT). The dosimetry associated with the study of such a physical change may be broken into three parts:

7.1.1 Determine the absorbed dose in a dosimeter.

7.1.2 Using the dosimeter measurement, estimate the absorbed dose in the region and material of interest in the DUT.

7.1.3 If required, relate the estimated absorbed dose in the DUT to the physical change of interest (holes trapped, interface states generated, photocurrent produced, etc.)

7.2 This section will be concerned with the first two of the above listed parts of dosimetry: (1) what is necessary to determine a meaningful absorbed dose for the dosimeter and (2) what is necessary to extrapolate this measured dose to the estimated dose in the region of interest. The final step in dosimetry, associating the absorbed dose with a physical change of interest, is outside the scope of this guide.

7.3 Energy Deposition:

7.3.1 Secondary Electrons—Both in the case of absorbed dose in the DUT and absorbed dose in the dosimeter, the energy is deposited largely by secondary electrons. That is, the incident photons interact with the material of, or surrounding, the DUT or the dosimeter and lose energy to Compton electrons, photoelectrons, and Auger electrons. The energy which is finally deposited in the material is deposited by these secondary particles.

7.3.2 Transport of Photons—In some cases, it is necessary to consider the transport and loss of photons as they move to the region whose absorbed dose is being determined. A correction for the attenuation of an incident photon beam is an example of such a consideration.

7.3.3 Transport of Electrons—Electron transport may cause energy originally imparted to electrons in one region to be

carried to a second region depending on the range of the electrons. As a result, it is necessary to consider the transport and loss of electrons as they move into and out of the regions whose absorbed dose is being determined. In particular, it is necessary to distinguish between equilibrium and non-equilibrium conditions for electron transport.

7.3.3.1 Charged Particle Equilibrium—In some cases, the numbers, energies, and angles of particles transported into a region of interest are approximately balanced by those transported out of that region. Such cases form an important class of limiting cases which are particularly easy to interpret. (See “Equilibrium Absorbed Dose” in Section 4.6.)

7.3.3.2 Dose Enhancement—Because photoelectron production per atom is roughly proportional to the atomic number raised to the fourth power for energies less than 100 keV (2), one expects more photoelectrons to be produced in high atomic number layers than in low atomic number layers for the same photon fluence and spectrum. Thus, there may be a net flow of energetic electrons from the high atomic number layers into the low atomic number layers. This non-equilibrium flow of electrons may result in an enhancement of the dose in the low atomic number layer. Dose enhancement problems are often caused by high atomic number bonding layers (e.g., gold), and metallization layers (e.g., W-Si or Ta-Si).

7.4 Absorbed Dose in Dosimeter:

7.4.1 Equilibrium Absorbed Dose in Dosimeter:

7.4.1.1 It is frequently possible to use dosimeters under approximately equilibrium conditions. The interpretation of the output of the dosimeter is straightforward only when the energy deposition processes within the dosimeter are approximately in equilibrium. That is, when the absorbed dose within the dosimeter is an equilibrium absorbed dose.

7.4.1.2 It is possible to treat non-equilibrium energy deposition within a dosimeter, but such an analysis requires electron and photon transport calculations, often in the form of computer codes.

7.4.2 Limiting Cases:

7.4.2.1 There are two limiting cases for which the dosimeter data can be analyzed in a straightforward manner.

7.4.2.2 Limiting Case One: Short Electron Range:

7.4.2.2.1 For this case, secondary electron ranges are small in comparison with the size of the dosimeter.

7.4.2.2.2 Essentially all electrons which deposit energy within the dosimeter will be produced within the dosimeter.

7.4.2.2.3 Non-equilibrium effects due to electron transport are negligible, but photon attenuation corrections may be necessary.

7.4.2.2.4 An example of this limiting case would be 20 keV photons depositing energy in a typical (0.889 mm thick) thermoluminescence (TL) dosimeter (TLD). In this case, the secondary electrons have ranges which are small in comparison with the size of the TL dosimeter. As a result, it is not necessary to perform a correction for the effect of electron transport on absorbed dose. On the other hand, 20 keV photons may be significantly attenuated while traveling through a TL dosimeter depending on the material. Thus a correction due to this effect may be necessary.

7.4.2.3 Limiting Case Two: Large Electron Range:

7.4.2.3.1 When the maximum secondary electron range is large compared with the size of the dosimeter, the dosimeter must be surrounded by an equilibrating layer. This layer must be chosen to be of an appropriate thickness, density and atomic number. Generally, the range of secondary electrons must be smaller than the thickness of the equilibrating layer. For further discussion of equilibrating layers, see Practice E 668.

7.4.2.3.2 Essentially all electrons which deposit energy within the dosimeter originate in the equilibrating layer.

7.4.2.3.3 Bragg-Gray cavity theory applies. That is, the dose within the dosimeter is the equilibrium dose for the equilibrating layer (corrected by the differences in electron stopping power of the dosimeter and the equilibrating layer and any photon attenuation through the material).

7.4.2.3.4 An example of this limiting case is 1-MeV photons incident on a typical TL dosimeter surrounded by an appropriate equilibrating layer. In this case, the range of the secondary electrons will be large in comparison to the size of the TLD. Thus the dose measured will be the equilibrium dose in the TLD (with a small correction for the differences in the stopping power for the electrons in the TLD material and the material of the equilibrating layer.)

7.4.2.4 Intermediate Cases

7.4.2.4.1 The intermediate cases, where secondary electron ranges are neither small nor large in comparison to the dosimeter size, are cases where non-equilibrium energy deposition is to be expected.

7.4.2.4.2 An example of an intermediate case is 100-keV photons incident on a typical thermoluminescence dosimeter.

7.4.2.4.3 The careful treatment of dosimetry for intermediate cases requires the use of combined photon/electron radiation transport calculations of the energy deposition in the dosimeters in order to get satisfactory measurements.

7.5 Absorbed Dose in Device Under Test:

7.5.1 Absorbed Dose in Device Under Test—The conditions within a DUT during a flash X-ray irradiation are frequently far from equilibrium. In many cases, these are classic examples of experiments showing dose enhancement effects. As a result, it is frequently necessary to perform a dose-enhancement correction in order to estimate the absorbed dose within the region of interest. Unless the DUT meets the two limiting cases discussed in the previous section, which is rarely true, the dose should be obtained using combined photon/electron radiation transport calculations of the energy deposition in order to get satisfactory dose estimates. Within a system being tested, a calculation of the radiation transport through the system to the individual devices of interest must be performed.

7.6 Spectral Considerations:

7.6.1 Broad Energy Range—The set of available flash X-ray sources spans a very wide photon energy range. Useful intensities may be obtained for energies as low as 10 keV and as high as 10 MeV. Each individual flash X-ray source, of course, does not produce useful photons over such a wide range. Dosimetry for such flash X-ray sources is somewhat simplified if the different flash spectra are categorized into three types, low energy, medium energy, and high energy. This categorization refers to the average energy of the photon spectrum and not the higher endpoint energy (Section 6.2).

TABLE 1 Flash X-ray Dosimetry Characteristics for Three Energy Bands

Flash X-Ray Type	Absorbed Dose in Dosimeter	Absorbed Dose in DUT
Low energy (average photon energy, 20-100 keV)	Can get electron equilibrium. ^A May need photon transport correction. ^B	Depends on DUT: May need an electron transport calculation. ^C May need a photon transport calculation. ^B
Medium energy (average photon energy, 100-300 keV)	Often cannot get electron equilibrium and therefore need electron transport calculation. ^{C,D}	Depends on DUT: May need an electron transport calculation, especially without use of beam filtration. ^{C,E} May need a photon transport calculation. ^B
High energy (average photon energy, 300-3000 keV)	Can get electron equilibrium with proper equilibration layer. ^{F,D}	Depends on DUT: May need an electron transport calculation. ^C Usually no photon transport calculation needed. ^D

^AThe dosimeter or region of interest is large compared to the maximum secondary electron range. (Section 7.4.2.2)

^BThe dosimeter or region of interest is large compared to the photon range. (Section)

^CThe dosimeter or region of interest is of comparable size to the maximum secondary electron range. (Section 7.4.2.4)

^DThe dosimeter or region of interest is small compared to the photon range. (Section)

^EA filter may be used to essentially eliminate the lower energy portions of the flash x-ray spectrum. This makes the spectrum more nearly monochromatic and may simplify dosimetry.

^FThe dosimeter or region of interest is small compared to the electron range. (Section 7.4.2.3)

7.6.2 *Dosimetry for Three Energy Types*—A summary of the dosimetry requirements for the three flash X-ray energy bands is provided in Table 1.

7.7 Absorbed Dose Interpretation:

7.7.1 In Section 7.1.2, it is pointed out that the second of three steps in the dosimetry process is the use of the measured absorbed dose in the dosimeter to estimate the absorbed dose in a region of interest within the device under test.

7.7.2 In the previous sections the dose in both the dosimeter (Section 7.4) and the DUT (Section 7.5) is estimated. Although the dose can easily be calculated from tables or a simple photon transport code for the dosimeter, which is often in equilibrium, the dose in the DUT, which is rarely in equilibrium, often requires a sophisticated combined photo-electron radiation transport code. Assuming this has been accomplished, one can use the ratio of the calculated doses at a given fluence and spectrum as the correction factor for the measured dose in the dosimeter to estimate the actual dose in the DUT.

$$D_{\text{DUT}} = \text{Calculated} \left(\frac{D_{\text{DUT}}}{D_{\text{Dosimeter}}} \right) \times D_{\text{Dosimeter}} \quad (5)$$

where D_{DUT} is the absorbed dose in the device material, and $D_{\text{Dosimeter}}$ is the equilibrium absorbed dose in the dosimeter.

8. Dosimetry Systems

8.1 *Introduction*—In this section a brief summary of eight different types of dosimetry systems is given. The intent of this guide is to provide enough information so that users can decide which system might be appropriate for their application. Users are expected to investigate in more detail the proper use and limitations of a system using references in this and other ASTM documents before undertaking any radiation measurements. The pros and cons of each system are quickly summarized in Table 2 and Table 3. The values cited in these Tables are indicative of what is typically obtained under normal testing conditions. Many of these parameters can be improved at the expenditure of considerable effort. Only the precision of each dosimeter is discussed here because the accuracy of the dosimeter will be determined by the quality of the calibration and specific usage and this is beyond the scope of this document.

TABLE 2 Dose Measurements

Dosimeter Type	TLD	Calorimeter	Optichromic	Radiochromic Film
Dose Range (Gy)	10^{-4} – 5×10^3	10 – 10^5	0.4 – 2×10^4	0.01 – 5×10^4
Photon Energy Range (MeV)	>0.01	>0.01	>0.01	>0.01
Size	Small	Medium	Medium	Small to Large
Ease of calibration	Easy	Easy	Easy	Moderate
Precision	5-15 %	5 %	5 %	5 %
Active Material	CaF ₂ , LiF	Au, Si, Al	Organic	Organic
Availability	Buy	Make or Buy	Buy	Buy
System Cost	Medium	Low	Low	Low to High

TABLE 3 Dose-Rate Measurements

Dosimeter Type	PIN Diode	Compton Diode	Cerenkov/Scintillator-Photodetector	PCD
Dose-Rate Range (Gy/s)	2×10^1 – 10^8	10^7 – 10^{10}	10^4 – 10^9	2×10^6 – 10^{10}
Photon Energy Range (MeV)	>0.01	>0.5	>0.02	>0.01
Size	Small to Medium	Large	Very Large	Small to Medium
Ease of calibration	Moderate	Difficult	Moderate	Moderate
Precision	10 %	10 %	4 %	10 %
Active Material	Si, GaAs	Tungsten	Plastic	Diamond, GaAs
Availability	Buy	Buy or Make	Make	Buy or Make
System cost	Low	Medium	High	Medium

8.2 Thermoluminescence Dosimeters:

8.2.1 *Introduction*—Thermoluminescence Dosimeters (TLDs), see Terminology E 170, are popular dosimeters because they are small, passive, inexpensive, and can retain accurate dose information for long periods of time between irradiation and readout. The dose range of TLDs is typically $>10^{-4}$ to $<5 \times 10^3$ Gy ($>10^{-2}$ to $<5 \times 10^5$ rad).

8.2.1.1 Many materials are available for thermoluminescence dosimetry; however, the favorites for radiation hardness testing are Lithium Fluoride (LiF), Manganese activated Calcium Fluoride (CaF₂:Mn), and Dysprosium activated Calcium Fluoride (CaF₂:Dy). Dosimeters are available as powder, chips made from polycrystalline material, and discs consisting of

very fine powder uniformly dispersed throughout a polytetrafluoroethylene (PTFE, Teflon) matrix. A commonly used size of chip is 3.2 by 3.2 by 0.9 mm (0.125 by 0.125 by 0.035 in.) A commonly used size of Teflon dosimeter is a disc, 6 mm in diameter and 0.4 mm thick. In addition, these dosimeter materials can be made into arrays for dose mapping.

8.2.2 Principles—Thermoluminescent materials consist of a crystalline insulator with added dopants which introduce stable electron traps into the forbidden band gap. Ionizing radiation creates electrons and holes which are trapped by stable traps in the band gap. The density of filled traps is proportional to the dose absorbed by the material. Subsequent heating of the material empties the electron traps, allowing electrons from F-centers to recombine with free holes at luminescence centers, emitting light. The integrated light output is proportional to the density of filled traps, and therefore to the absorbed dose in the TLD material.

8.2.2.1 To make an equilibrium dose measurement, the TLD must be enclosed in an equilibrium capsule of the appropriate material and thickness when it is exposed. Methods for determining equilibrium capsule thickness are given in Practices E 665 and E 666.

8.2.2.2 Readout of TLDs is accomplished by an instrument consisting of a heater, optical system and photomultiplier detector to measure the light emitted by the TLD during a predetermined heating cycle, and an integrating picoammeter which can measure the current or the charge from the photomultiplier. The total integrated charge from the photomultiplier during part or all of the heating cycle is usually related to the absorbed dose in the dosimeter. TLD readout instruments are available from a number of manufacturers.

8.2.3 Applications—The choice of the TLD and capsule material depend on the application. For high-energy photon sources ($h\nu \geq 200$ keV) such as high voltage flash X-ray generators and Co^{60} and Cs^{137} sources, and where the quantity of interest is Si or SiO_2 equilibrium dose, a good combination of materials is a $\text{CaF}_2\text{:Mn}$ TLD in an Al equilibrium shield (See Practice E 1249). For these materials and photon energies, all mass energy absorption coefficients and mass stopping powers are so close to those of silicon that the silicon equilibrium dose can be determined using Bragg-Gray cavity theory. The response of $\text{CaF}_2\text{:Mn}$ is nearly linear with absorbed dose and the dosimeters retain dose information for a long time with a small fading correction, whose form is well known (8.2.5).

8.2.3.1 For radiation sources which have significant photon energy below 200 keV, more detailed calculations are necessary regardless of which material is chosen for the TLD. In this case the choice of TLD and capsule material is flexible, and can be made on the basis of convenience or cost.

8.2.3.2 The choice of whether powder, chips, or PTFE discs are used is primarily one of convenience and cost. For specific instructions on the proper procedures for using TLD systems to determine absorbed dose in radiation hardness testing of electronic devices, see Practices E 665, E 666, and E 668.

8.2.4 Advantages—TLDs are small, inexpensive, and require no instrumentation during irradiation. They are thus ideally suited for measuring dose at many locations within a test object, or for measuring dose near the areas of interest in

even very small (~ 3 mm) test objects. Their large sensitivity range makes it possible to measure Gy to kGy doses with one dosimetry system.

8.2.5 Limitations—Energy Range: TLDs are usable over the energy range from at least 10 keV to 10 MeV. (Indeed, their sensitivity extends down into the ultraviolet region; hence, TLDs must be protected from exposure to sunlight, etc., especially for low dose applications.) The dose response as a function of energy can be calculated for TLDs which consist of 100% TL phosphor by using standard equilibrium dose and Bragg-Gray cavity theories. (Practices E 665 and E 666) However, care must be exercised when using PTFE disc dosimeters at low energies. Since these dosimeters consist of fine grains of TL phosphor distributed throughout a PTFE matrix, these simple theories do not apply. At high energies ($h\nu \geq 2$ MeV) the TLD can be treated using Bragg-Gray cavity theory, with all photon interactions assumed to take place in the equilibrium shield material, and the resultant secondary electrons depositing energy in the phosphor grains. There is a smooth transition to more moderate energies ($h\nu \approx 200$ keV) where the photon interactions must be considered to occur primarily in the PTFE matrix. Finally, at the lowest photon energies ($h\nu \leq 10$ –20 keV), all photon interactions which eventually deposit energy in the phosphor grains must be assumed to originate in the individual phosphor grain itself. The dose response of PTFE TLDs as a function of photon energy is therefore a complex function of energy which is not easy to derive analytically. A good electron photon transport code such as TIGERP (3) can be helpful in calculating this relationship (4).

8.2.5.1 Most TLD phosphors exhibit fading of the TL response with time; that is, the observed TL response becomes progressively smaller as the time interval between irradiation and readout increases (5). For most of the common TL phosphors the fading behavior is well known. The fading of $\text{CaF}_2\text{:Mn}$ TLDs has been reported to obey the relationship (6)

$$(\text{TL})/\text{TL}_0 = (t/t_0)^{-0.017} \quad (6)$$

where t is in hours and the subscript 0 refers to the time of irradiation, and unsubscripted quantities refer to the time of readout. The value of the coefficient 0.017 is only typical and can vary from batch to batch and must be determined experimentally. Use the procedures in 8.7 of Practice E 668 to test for fading effects in the type of TLD chosen.

8.2.6 Sensitivity—TLDs can be used to measure dose from 100 μGy to 5 kGy (10 mrad to 500 krad). Most TL materials saturate in the range of 3–10 kGy (300 krad to 1 Mrad). The dose response of calcium fluoride TLDs is independent of dose rate for dose rates less than 10^{10} Gy/s (10^{12} rad/s) (7). The dose response of lithium fluoride has been reported to be independent of dose rate only up to 10^9 Gy/s (10^{11} rad/s) (8).

8.2.7 Calibration—TLD systems (dosimeter plus reader) must be calibrated in a standard radiation field before use, see ICRU Report 14. The most convenient sources for this are calibrated Co^{60} or Cs^{137} sources. Care must be taken to expose the TLDs in an appropriate equilibrium shield. If the calibration source is calibrated in units of exposure rate, then the dose absorbed by the TLD can be found by:

$$D_{\text{TLD}} = \left(\frac{(\mu_{\text{en}}/\rho)_{\text{TLD}}}{(\mu_{\text{en}}/\rho)_{\text{air}}} \right) (D_{\text{air}}/X) X_0 e^{((\mu_{\text{en}}/\rho)_{\text{air}} - (\mu_{\text{en}}/\rho)_{\text{TLD}}) X_0} \quad (7)$$

where:

- D_{TLD} = is the dose absorbed by the TLD,
- X_o = is the free-in-air exposure of the source,
- D_{air}/X = 33.68 Gy·kg/C (0.869 rad/R),
- μ_{en}/ρ = is the mass-energy absorption coefficient,
- ρ_{eqmat} = is the density of the equilibrator material, and
- x = is the thickness of the equilibrator material.

8.2.8 Reproducibility—The reproducibility of most TLD materials is about $\pm 8\%$ at 1σ . Better reproducibility can be achieved with TLD chips by irradiating a number of them to a low dose, reading them, and selecting those which fall into a tighter group. A reproducibility of $\pm 2\%$ is often achievable, and $\pm 1\%$ is possible by this means.

8.3 Calorimeters:

8.3.1 Introduction. Calorimetry comes closest of all the dosimetric techniques to providing a direct measure of dose or fluence. Apart from corrections for thermal leakage, thermal defect due to chemical or solid state reactions, and energy loss in fluence measurements, only the specific heat and the temperature rise of the absorber due to X-ray absorption need be known. The specific heat or the heat capacity of the absorber is a thermodynamic property of the given material. Temperature rise measurements can be made with thermocouples, thermopiles, resistive temperature detectors (RTDs), thermistors, etc.

8.3.2 Principles—The temperature rise of the absorbing material is related to the energy absorbed, enthalpy or dose, in the material through its specific heat. Namely,

$$\Delta H = \int C_p dT \quad (8)$$

It has been assumed that the absorber remains solid at constant

pressure and that no phase transition or other process occurs which would complicate this simple relationship.

8.3.2.1 The specific heat (9), that is, the heat capacity per unit mass for a typical absorbers such as tantalum varies from 139.7 to 141.6 J·kg⁻¹·K⁻¹, or for aluminum from 900.6 to 919.0 J·kg⁻¹·K⁻¹ between 20°C and 60°C. Hence, for many materials over practical temperature changes, a constant specific heat can be used provided accuracies better than 1 % are not required. For improved accuracy the specific heat may be expanded in a power series in temperature over the temperature range of interest.

8.3.2.2 Temperature sensors such as thermocouples (10) are used routinely by properly attaching the thermocouple wires to the absorber. An example of a thermocouple attached to an absorber is shown in Fig. 2. The 0.025-mm thick gold foil is thermally isolated from the surrounding environment by supporting the foil on small nylon threads and by using small diameter thermocouple wires. The thermocouple wires pass through a lead shield and then they are soldered to miniature connector pins with lead free solder. The lead shield and lead free solder are used to reduce the X-ray induced temperature rise in the solder connection. In order to improve charged particle equilibrium and reduce fluorescence losses from the gold, a gold backing foil is placed just behind the gold absorber. A coupled photon/electron transport code should be used to determine how far this geometry is from achieving charged particle equilibrium for a particular spectrum. Although not shown in the figure, an optional fine wire can be welded to the gold foil. When this is done, the calibration of the

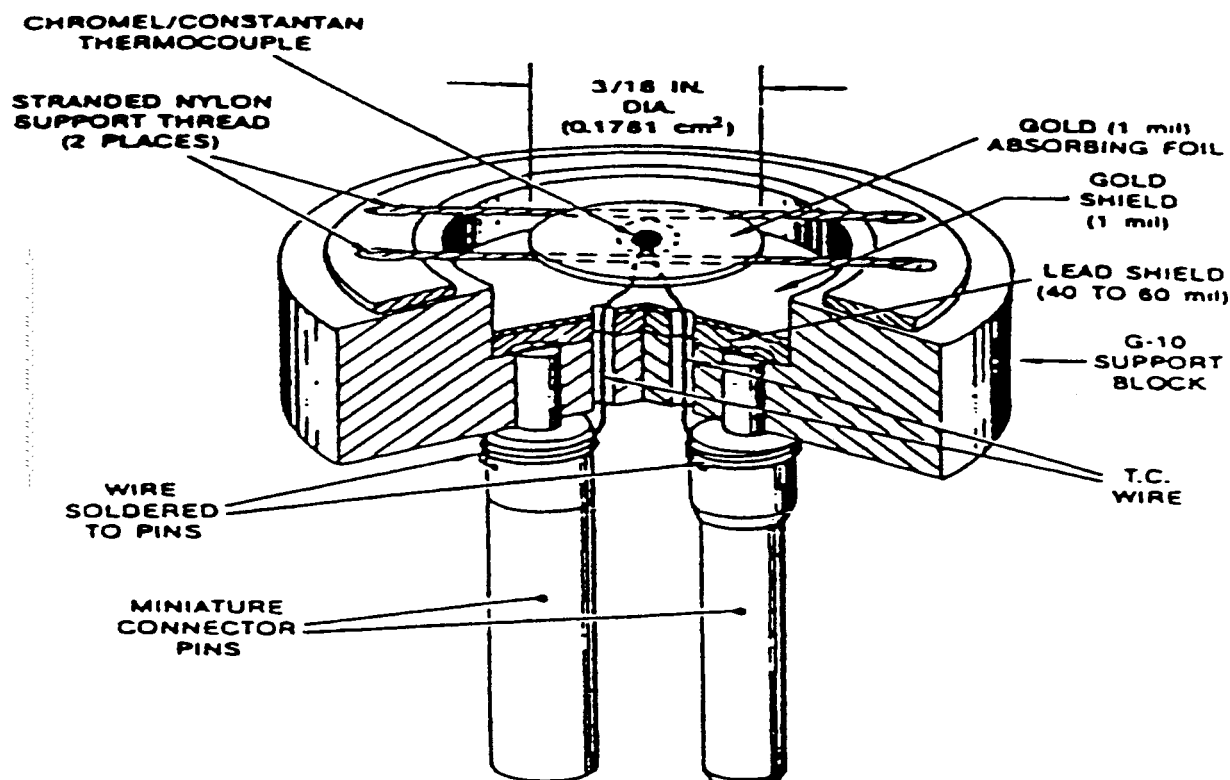


FIG. 2 Cutaway View of a Typical Gold Foil Dose Calorimeter Designed for Use with FXR Spectra Which Have Maximum Energies Less Than 2 MeV (courtesy of Maxwell Laboratories, Inc., San Diego, CA).

calorimeter can be checked with a proton Van de Graaff provided the thermocouple is electrically isolated from the X-ray absorber.

8.3.2.3 Thin foil X-ray absorbers are typically designed for measuring dose in the absorbing material. A thicker absorber may be used as a total fluence detector for low energy spectra. However, in this case a more sensitive temperature detector such as a thermistor may be required and a much thicker absorber. A typical thermistor has a negative temperature coefficient of resistivity, α , which decreases in magnitude from about 0.039 to 0.036/°C. This coefficient is defined by:

$$(\Delta\rho)/\rho_o = \alpha(\Delta T) \quad (9)$$

where ρ_o is the original resistivity and $\Delta\rho$ and ΔT are the change in resistivity and temperature. Hence, care must be taken to measure the temperature both before and after X-ray energy absorption. The temperature sensitivity of a thermistor may be closely approximated with the Steinhart-Hart equation (11),

$$1/T = A + B(\ln R) + C(\ln R)^3 \quad (10)$$

where T is in K and R is the thermistor resistance. The values of A, B, and C should be measured for each thermistor and these values should be remeasured systematically to ensure that the thermistor has not been damaged.

8.3.3 *Advantages*—Calorimeters can be made absolute, either intrinsically or by means of electrical-heating calibration.

8.3.3.1 The measurement of temperature rise comes closest of any dosimetric technique to being a direct measurement of the energy involved in the absorbed dose.

8.3.3.2 Calorimeters are inherently dose-rate-independent under adiabatic conditions, and become more convenient to use as the dose-rate increases because thermal leakage during dose delivery becomes negligible.

8.3.3.3 Metal film calorimeters have no LET dependence (neglecting minor differences in thermal defect, if any), since ionic recombination is irrelevant to the temperature rise.

8.3.3.4 The conversion of absorbed dose to a temperature rise takes place on the order of a few picoseconds. Hence, the temperature rise in the material for current FXRs follows the running integral of the dose rate, and calorimeters can be designed to measure the true absorbed energy in a small region or the average dose in a larger volume after thermal equilibrium.

8.3.4 *Limitations* —(12). The temperature rises to be measured are typically small and in many cases are only a fraction of a degree. This limits calorimetry to relatively large doses.

8.3.4.1 The calorimeter absorber must be designed to allow the measurement of the temperature rise before excessive thermal losses takes place. This requirement must be considered in conjunction with requiring charged particle equilibrium.

8.3.4.2 The recording of the calorimeter signals are different than most of the other signals associated with FXR tests and, consequently, the test has an additional complication when calorimeters are used. In many cases calorimeter signals are a few tens of microvolts to a few millivolts and must be recorded with a few hundred hertz frequency response. In these cases high-input-impedance, high-gain amplifiers in conjunction with low pass filters are often used.

8.3.4.3 In some instances the dose from extremely high fluences from lower energy spectra can be high enough to either melt the absorber or, for slightly lower dose values, generate a thermomechanical shock which may break the absorber, the temperature sensor, or perhaps, detach the temperature sensor from the absorber.

8.3.5 *Sensitivity*—Apart from the sensitivity of the temperature sensor and the specific heat of the absorber, the basic sensitivity of the dose calorimeter is dependent upon the mass energy absorption coefficient of the X-ray absorber. The degree to which the absorber deviates from this ideal cross section must be calculated with a code such as the ITS (3). A similar comment can be made for total-fluence calorimeters except that in this case the deviation from total incident energy absorption would be calculated.

8.3.6 *Calibration*—The output of a properly designed calorimeter can be interpreted from the intrinsic knowledge of the specific heat of the absorbing material and the calibration of the temperature sensor. Alternatively, several techniques are available to confirm that the calorimeter has been properly designed. Examples of these techniques include embedded electric heaters, proton Van de Graaff pulse heating, and flash lamp pulse heating. The temperature sensor element should be checked periodically for changes in response over time.

8.3.6.1 Besides the basic calorimeter calibration, the calibration of the high gain recorder system should be checked. It is recommended that a step voltage pulse be used to check the gain of the system on each channel every day during a test series in which the calorimeters are being used. The step pulse should be applied physically in place of the calorimeter so that the entire signal line is included in the calibration.

8.3.7 *Reproducibility*—Apart from electrical noise and the basic calibration of the specific heat and temperature detector, the reproducibility and accuracy of the dose or fluence measurement are limited by the readability of the recorded signal. In some cases this may be dominated by electrical noise, thermal noise, or heat exchange. As a practical figure of merit, the reproducibility for FXR measurements should be about 5 %.

8.4 Opti-chromic Dosimeters

8.4.1 *Introduction*—Opti-chromic Dosimeters (ODs) (13-18) are a relatively new type of dosimeter that has many of the same advantages and uses of TLDs. They are relatively small (~3-mm diameter and 25 - 50-mm long), passive, inexpensive and retain accurate dose information for long periods of time (months) between irradiation and measurement of dose. The useful dose range of the ODs is >0.4 Gy to <20 kGy (>40 rad to <2 Mrad).

8.4.1.1 Organic solvents with a high refractive index are used to fill a hollow fluorinated plastic tube having a low refractive index, forming an optical waveguide. Radiochromic dye is dissolved in these solvents and sensitized by them. Glass beads are located in the ends of the tube and serve as lenses for the waveguide. Dosimetry is performed by measuring changes in the optical density of the fluid.

8.4.2 Principles—The dosimeters are supplied as fluorinated polyethylene-polypropylene (FEP) tubing (50 or 25 mm long, 3 mm o.d., wall thickness 0.3 mm) filled with hexahydroxyethyl pararosaline cyanide (HPC) dissolved in a mixture containing triethyl phosphate, dimethyl sulfoxide, and polyvinyl butral. The filled tubing is sealed at both ends with glass beads (diameter 3 mm) forming a waveguide

8.4.2.1 The HPC is a colorless pre-cursor of a common highly colored stable organic dye. When a liquid solution of this compound, in these polar solvents, is irradiated with ionizing radiation having an energy exceeding ~ 4 eV, the cyanide group is split off. This results in an electron rearrangement in the parent molecule which yields a blue dye. In this solution, the resulting color change is very stable. The change in the optical density is a monotonic function of the absorbed dose in the solution.

8.4.3 Applications—These dosimeters can be used for virtually all applications where TLDs are used. They are somewhat easier to store, handle and read. They cannot be used for very low doses nor for dosimetry over very small areas, because of their large size in comparison to TLDs. For high-energy photon sources ($h\nu \geq 300$ keV) such as high-voltage flash X-ray generators and Co^{60} and Cs^{137} sources, a good choice of additional equilibrium shield is plastic. For these high energies and materials the mass absorption coefficients and mass stopping powers are so close to those of silicon that the silicon equilibrium dose can be determined using simple Bragg-Gray cavity theory. For radiation sources which have a significant part of the energy spectrum below 300 keV, the response of the fluid is akin to that of water and more detailed calculations are necessary to obtain dose silicon.

8.4.4 Advantages—This system can be used for many of the applications for which TLDs are used and is a passive diagnostic. The reader is simple, portable and requires little maintenance. The response of the opti-chromatic dosimeters is nearly linear with dose. In addition, readings can be made a few minutes after the irradiation and there is no fade correction for a period of months for doses < 2 kGy.

8.4.5 Limitation, Physical—The complete dose range of 0.4 Gy to 10^4 Gy is not spanned by a single OD. The concentration of HPC determines the sensitivity of the dosimeter. A typical range of use for any concentration is 2 orders of magnitude in dose. Although this is a wide range of doses, one must be able to predict the dose at least to this degree so that a dosimeter with the appropriate concentration can be selected.

8.4.5.1 The chemical reaction is slightly temperature sensitive. Over the range of 0° to 60°C the response has a temperature dependence of $\sim 0.2\%$ per $^\circ\text{C}$ (13, 18). For normal irradiation conditions this is usually insignificant.

8.4.5.2 The dosimeters must be kept refrigerated when stored for long periods of time to prevent “aging” of the solution. If this is done the dosimeters have an effective shelf-life of ~ 1 year. Recalibrations should be performed about every 6 months.

8.4.5.3 The fluid is sensitive to ultraviolet light. Care should be taken to minimize the exposure of the dosimeter to natural or fluorescent lighting for significant periods of time (> 1

minute). Exposure to ultraviolet light will darken the fluid and yield abnormally high dose readings.

8.4.5.4 Energy Range—ODs are usable over the energy range from at least 10 keV to 30 MeV. The dose response can be calculated as a function of energy for the fluid using standard equilibrium dose and Bragg-Gray cavity theories. (Practices E 665 and E 666.) For lower energies, the dosimeters will give doses significantly different than for silicon. In addition, the thickness of the dosimeter and the different materials (FEP, Al) used to surround the dosimeter may result in a non-equilibrium and/or non-uniform dose in the OD. These possible effects can be calculated using a good electron-photon transport code such as ITS or CEPXS/ONELD.

8.4.5.5 Time Dependence—There is no significant fade or time dependence in the optical density after irradiation for doses < 2 kGy. Above this dose there is a change in the optical density of the glass beads in the ends of the dosimeter which reaches a steady state value after a few days. This time dependence must be considered when using or calibrating these dosimeters at very high doses.

8.4.6 Sensitivity—ODs can be used to measure dose from 0.4 to 20 kGy (40 rad to 2 Mrad) using a number of different HPC concentrations. The dose response of the OD is independent of dose rate up to 10^{12} Gy/s (10^{14} rad/s) (16).

8.4.7 Calibration—An approved method for using optical waveguide dosimetry is given in Practice E 1310. In order to make a valid equilibrium dose measurement, the OD must be enclosed in an appropriate material and thickness when it is exposed. Methods for determining the proper equilibrium capsule thickness are given in Practices E 665 and E 666. Practice E 1310, Section 7 provides calibration procedures traceable to national standards.

8.4.7.1 The irradiated dosimeter is placed in a spectrophotometer that uses light of narrow wavelength bands at several wavelengths to determine radiation-induced increases in the optical density of the dye. Optical densities at those wavelengths are referenced against a base wavelength where the optical density of the dye is insensitive to radiation. For increased precision, individual preirradiation optical density readings can be subtracted from the post irradiated values.

8.4.8 Reproducibility—The reproducibility of ODs is about 5 % at 1σ over the entire dose range when preirradiation measurements are taken for each dosimeter. The calibration will drift with time and should be checked at least every 6 months.

8.5 Radiochromic Film Dosimeters:

8.5.1 Introduction—Radiochromic film dosimetry utilizes a radiation induced photochemical reaction to determine absorbed dose. The change in optical density of the radiochromic material is correlated to the dose. The optical density is measured using a simple photo-optical system. The films are easily calibrated and generally have an extensive absorbed dose range (see Practice E 1275). Certain types of films can be used for irradiations greater than 100 kGy (19, 20).

8.5.2 Principles—Radiochromic materials are fabricated in either a solid state solution as plastic films or as a gel coating on a plastic substrate. The solid state form contains a dye

precursor which initially is transparent and colors upon exposure to either ultraviolet light or ionizing radiation. A dye derivative which is commonly used is hexahydroxyethyl aminotriphenyl aceto-nitrile. This colorless derivative changes to a deeply colored state as a function of absorbed dose. Gel coatings consist of a thinly-coated radiation sensitive layer on a plastic substrate material such as polyester. The change in color for both types of radiochromic media can be measured using a simple photometric device.

8.5.2.1 The optical density of radiochromic material varies with wavelength. The response curve is bell shaped, having a peak at a particular wavelength that is dependent on the radiochromic material used. At higher doses, the response ultimately saturates at the peak optical density and the curve begins to flatten.

8.5.3 *Applications*—Radiochromic films have a variety of uses as a dosimeter material. Depending on the radiochromic material, absorbed doses can be measured with certain materials having dose sensitivity that ranges from 0.01 Gy to 50 kGy. Advances in film for medical purposes have extended the dose sensitivity down to 0.01 Gy. Additionally, thin film dosimetry is a very useful tool for measuring dose gradients and electron or photon beam profiling (21, 22). In addition, because the film is often thin with respect to the secondary electron range, the dosimeter can be made to measure the dose in the material of choice using an appropriate equilibrater.

8.5.4 *Advantages*—Measuring the optical density of radiochromic films does not affect the transmission characteristics of the film. Consequently, the absorbance of any film can be measured an unlimited number of times.

8.5.4.1 After the response has reached its peak value from irradiation, the absorbance does not change appreciably for several months. There is essentially no fading associated with radiochromic films. Because it is a film, it is ideal for dose mapping. Although it has been common in the past to use a densitometer, these days, it can be as simple as using an appropriate filter and a flat bed scanner to obtain a dose map from radiochromic films.

8.5.4.2 The response is approximately linear with absorbed dose up to a saturation level and generating a calibration curve for films is very straightforward.

8.5.5 *Limitations*—Some radiochromic materials can show variations in response due to differences in temperature, dose rate, and humidity. The reproducibility from one batch to another may vary due to differences in thickness and concentration of the radiochromic chemical within the dosimeter matrix (23).

8.5.5.1 *Absorption Spectrum and Dose Response*—The range of wavelengths at which the optical density can be measured varies depending on the radiochromic material. In virtually every case there is a particular wavelength that generates a maximum absorbance. At a certain value of absorbed dose, the peak response will saturate. Greater doses can be measured, however, by shifting to an off-peak wavelength (24).

8.5.5.2 *Post-irradiation Stability*—The color after irradiation does not develop immediately in radiochromic films. Generally, there will be a delay in the color build up for most

types of films. The delay time associated with any change in absorbance from irradiation can range from a few minutes to a few days depending on the material (25, 26).

8.5.5.3 *Dose Rate*—Studies with several types of radiochromic films have demonstrated no dose rate dependence on response. This has been shown for dose rates up to 10^{12} Gy/s (27).

8.5.5.4 *Temperature During Irradiation*—Because a photochemical reaction is taking place, radiochromic thin films will be affected by the temperature during an irradiation. The functional dependence of response on temperature will depend on the type of radiochromic material. Therefore, an appropriate correction factor is necessary in determining dose response for a given irradiation temperature (28).

8.5.5.5 *Relative Humidity*—Some types of radiochromic thin films have a distinct humidity dependence on the response. The humidity effects of a radiochromic nylon film, for example, will be negligible between 35 %-65 % RH. Outside the limits of this range, however, the response falls off significantly.

8.5.5.6 *Light Sensitivity*—Radiochromic films become colored when exposed to radiation with a wavelength of 350 nm or less. Consequently, sunlight and fluorescent light will color radiochromic films and procedures should be implemented to minimize this effect.

8.5.6 *Sensitivity*—For the majority of radiochromic thin films available, the response is a linear function of absorbed dose and the sensitivity per unit thickness does not vary with the thickness of the films. However, thickness non-uniformity across a film requires compensation. A calibration curve per unit thickness taken from one thickness of film is generally applicable to all thicknesses. However, the calibration of a particular type of radiochromic film is batch dependent.

8.5.7 *Calibration*—Dosimeters may be calibrated in several ways such as in a national calibration facility, in an in-house local facility whose dose rate is traceable to national standards, or in the end-use radiation facility along with appropriate reference or transfer standard dosimeters (see procedures in Section 8 of Guide E 1261).

8.5.8 *Reproducibility*—For commercially available thin radiochromic films, the standard deviation of dose measurement from a single dosimeter is within ± 5 %. Most of the variation is due to differences in film thickness. Reproducibilities of ± 2 % or better can be obtained by taking multiple dosimeter measurements and taking a mean optical density or by measuring individual film thickness and calculating a response per unit thickness.

8.6 PIN Diode Dosimeters:

8.6.1 *Introduction*—A PIN diode collects the ionization produced in its intrinsic region to produce a current proportional to the deposited dose rate. The system is relatively straightforward to implement in its simplest form requiring only materials available at most flash X-ray facilities. The primary drawback is saturation of the signal at dose rates typical of flash X-ray facilities.

8.6.2 *Physical Principles*—The PIN diodes may be viewed as a solid-state ion chamber. Reversed-biasing a diode creates a depletion (or intrinsic) region sandwiched by p and n regions.

Electron-hole pairs generated in the depletion region are separated by the applied potential difference and appear as a reverse current sourced by the diode. The current is proportional to the rate of deposited dose.

8.6.2.1 The size of the intrinsic region in the PIN diode is dependent more on the diode construction than on the reverse bias. Thus, the sensitive volume is only a weak function of the applied voltage, making the PIN diode relatively insensitive variations in voltage. This is a major advantage for the PIN diode, especially as the induced current tends to reduce this bias.

8.6.2.2 The electron and hole mobilities in the diode material are sufficiently high as to not be a factor in the time response of the diode. The intrinsic capacitance of the diode (29), as well as the reactance of the cabling and detector system, serve to slow down the overall response. Rise times of the order of 1 ns may be obtained.

8.6.2.3 PIN diodes are constructed with a variety of materials, so it is often difficult to understand the exact nature of the response to radiation. In this case, they are typically calibrated by comparison to a better-controlled dosimeter, such as a TLD. Once calibrated, the diode may be used repeatedly in the same radiation field. PIN diodes specially constructed as Bragg-Gray cavities are available.

8.6.3 *Applications*—PIN diodes provide the capability to obtain dosimetry information on-line, without entry to the flash X-ray cell or opening up a large experimental configuration. When located at fixed distances from a flash X-ray machine, these detectors perform well as dose-rate and timing monitors. PIN diodes may also be employed as a flash X ray output diagnostic or for field mapping. Large PIN diodes (100 cm²)

can be obtained for very low dose rate applications. Because of saturation, PIN diodes are of less use in the region of maximum beam intensity (30, 31).

8.6.3.1 Correct electronic configuration of the diode dosimeter is essential for the proper functioning of the system. Fig. 3 shows a typical setup. The biasing capacitor may be located close to the diode to maintain biasing voltage during the flash X-ray pulse. Sometimes two capacitors are used: a larger to supply the bulk of the charge, and a smaller, low-inductance capacitor for rapid response. The bias voltage supply is decoupled from the circuit during the pulse by a resistor. Since many of the components of the PIN dosimetry system are exposed to at least part of the flash X-ray radiation environment, their response to the radiation may form part of the system response. A useful technique to help separate the diode response from noise is to expose the system to flash X-ray pulses with all components in place except for the diode. The diode is replaced either with an open circuit (such as using an epoxy insulator) or with a short circuit. With the open circuit, the system may be run under bias. Signals obtained with this setup are noise, and must be maintained small compared to the signal obtained with the diode in place.

8.6.4 *Advantages*—PIN diodes have the advantages of small size, low cost, and high sensitivity. Another advantage to using diodes as dosimeters is that the diode material may be selected to match the material of interest. For example, PIN diodes of silicon and gallium arsenide are both available. However, one must be extremely careful that the material surrounding the diode does not interfere with the desired response and that electron equilibrium is obtained.

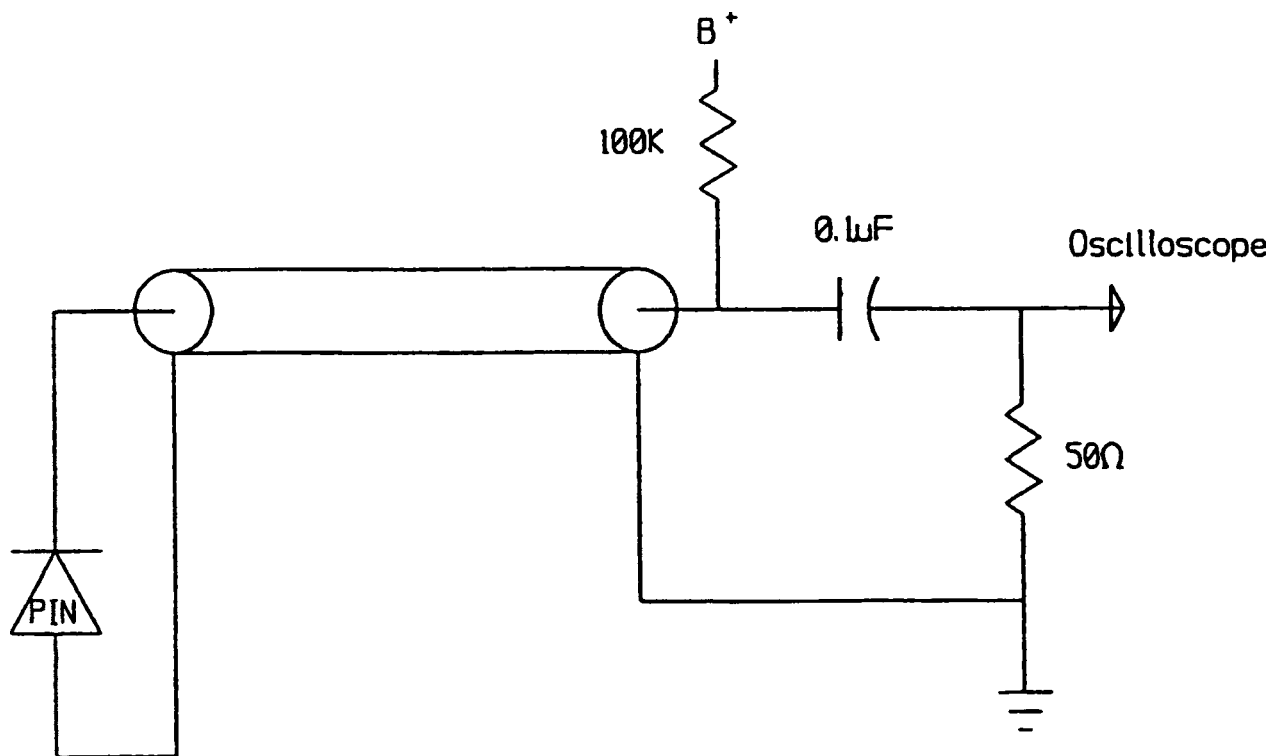


FIG. 3 Typical PIN Diode Configuration

8.6.5 Limitations—PIN diodes easily saturate at values typical of flash X-ray testing. This comes about from debiasing of the voltage supply and from recombination of electron–hole pairs in the intrinsic region. Also, a PIN diode cannot source current to a higher voltage than its supply. For example if a PIN diode using a 200 V bias produces a 150 V signal at dose rate X , it cannot produce a 300 V signal, exceeding the 200 V bias, at dose rate $2X$. These effects combine to distort the pulse shape.

8.6.5.1 The low dose-rate limit on the use of a PIN diode is caused by reverse-current leakage. The relative effect of this may be reduced by using a larger-volume diode. Cooling the diode is another, but seldom-used, possibility for reducing the leakage current.

8.6.5.2 Off-the-shelf diodes are often of unknown construction and can give rise to a variety of attenuation, dose enhancement and build-up effects. Hence, it may still be necessary to know the radiation spectrum and use a radiation transport code to infer the equilibrium dose or the dose in the DUT, even if the PIN is made of the same materials the DUT.

8.6.6 Sensitivity—PIN diodes are available for measuring dose rates from below 2×10^1 up to 10^8 Gy(Si)/s (2×10^3 up to 10^{10} Rad(Si)/s) with rise times as short as 1 ns. At dose-rates beyond the saturation level, the diode underresponds. Calibration against dose-rate insensitive dosimeters must be used to find the appropriate working range for each diode.

8.6.6.1 The sensitive volume of a diode will have a covering layer of doped semiconductor as well as a means of charge collection. In high energy fields, care must be taken to ensure that the cover is thick enough, and of the correct material, to provide electron equilibrium. In low energy fields, attenuation, dose enhancement, and dose build-up may all complicate the diode response (32).

8.6.7 Calibration—The calibration of the diode detector may best be performed by observing its behavior in a beam of known intensity and time variation. The intensity should be higher than expected in the flash X-ray environment, and the time variation faster. A LINAC may provide such a field.

8.6.7.1 Alternatively, the diode response may be integrated over time to find the total delivered dose. This dose is then compared to a total-dose detector (such as a TLD) exposed simultaneously with the diode. Saturation is revealed by a falling-off of the integrated diode response as compared to the total-dose detector at high dose rates. Should saturation occur at undesirably low dose rates, a higher bias voltage may be considered, subject to diode breakdown.

8.6.7.2 The calibration of the diode may also be determined by considerations of diode size and material. The dose rate deposited in an ideal diode is given by (33):

$$\dot{D} = 1.602 \times 10^{-14} \frac{IW}{eSL\rho} \quad (11)$$

where

- I = is the induced current (amps) in the depletion layer,
- W = is the average energy (eV) to produce an ion pair (3.6 in Si, 2.8 in Ge, 4.8 in GaAs),
- e = is the electron charge 1.602×10^{-19} C,
- S = is the depletion region surface area (cm^2),

- L = depletion layer thickness (cm),
- ρ = semiconductor density (g/cm^3), and (2.33 for Si, 5.32 for Ge, 5.32 for GaAs).

8.6.7.3 Radiation damage can alter the diode response. Generally, the carrier lifetime should be maintained an order of magnitude larger than the charge collection time. Neutron exposures of 10^{12} n/cm² or gamma-ray exposures of 10^5 cGy(Si) [rad(Si)] may be sufficient to cause improper response. Detectors exposed to these levels should be periodically recalibrated.

8.6.8 Reproducibility—Operating in the linear range, PIN diodes can measure dose rates with a precision of better than 5 %. The main limiting factor is electrical noise introduced in the cabling.

8.6.8.1 Biases in the result will be introduced by the calibration procedure. Diode dimensions may not be exactly known, or a reference dosimeter used to calibrate the diode may have its own bias. Measurements to within 10 % of true values may be obtained.

8.7 Compton Diodes:

8.7.1 Introduction—Dose rates for intermediate- to high-energy photon sources (>0.3 MeV) can be measured by Compton diodes (CDs), which give an electrical signal proportional to the rate of production of secondary electrons (especially Compton electrons) by incident radiation (34-36).

8.7.2 Principles—They are simple in design, consisting of an inner and outer conductor with either a dielectric or vacuum between. The dielectric-filled CD (37-40) has a thin front window, a low-Z oil or solid dielectric in which forward-directed Compton electrons are generated, and a high-Z collector which collects the electrons and directs them to the signal cable (Fig. 4). The collected electrons produce a negative signal proportional to the dose rate. By contrast, the vacuum CD (41) has a thin front window separated by a vacuum region from an intermediate- thickness aluminum emitter plate (Fig. 5). The partially-absorbed incident radiation generates Compton electrons which escape the emitter plate, producing a positive signal proportional to dose rate. The use of a DC magnet in front of the vacuum CD is recommended to sweep out any incident electrons, which would distort the signal.

8.7.3 Advantages—Compton diodes are low cost, require no bias, and are mechanically rugged. Compton diodes have been shown to have a response proportional to dose rate from ~ 0.1 Gy/s to $>10^{10}$ Gy/s (~ 10 rad/s to $>10^{12}$ rad/s) (41). Signals >1000 V have been observed at higher dose rates, finally limited by electrical breakdown in the CD or signal cable (42). For some fiducial signal generation applications, CDs with high voltage connectors and larger diameter collectors have been designed to produce signal amplitudes of several kV (38). Generally, signal amplitude should be proportional to the collector (emitter) area, except for edge effects which can be eliminated by collimating the incoming radiation beam.

8.7.4 Limitations—Spectral Sensitivity: Compton diodes show a substantial spectral sensitivity (possibly including signal polarity reversal) for low-energy photons (<0.1 MeV),

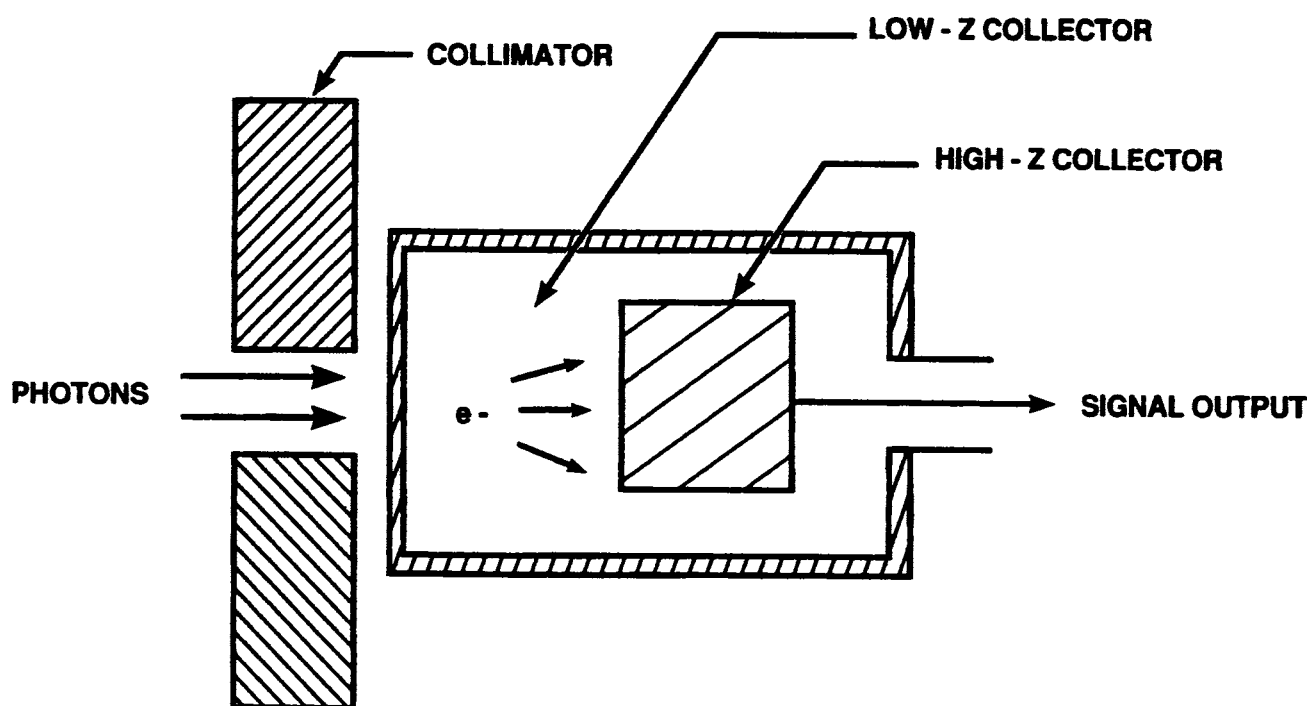


FIG. 4 Dielectric-Filled Compton Diode Schematic. Photons interact with the dielectric filling, creating secondary electrons (e^-) which are stopped in the collector, producing an output signal (-) proportional to the rate of energy deposition.

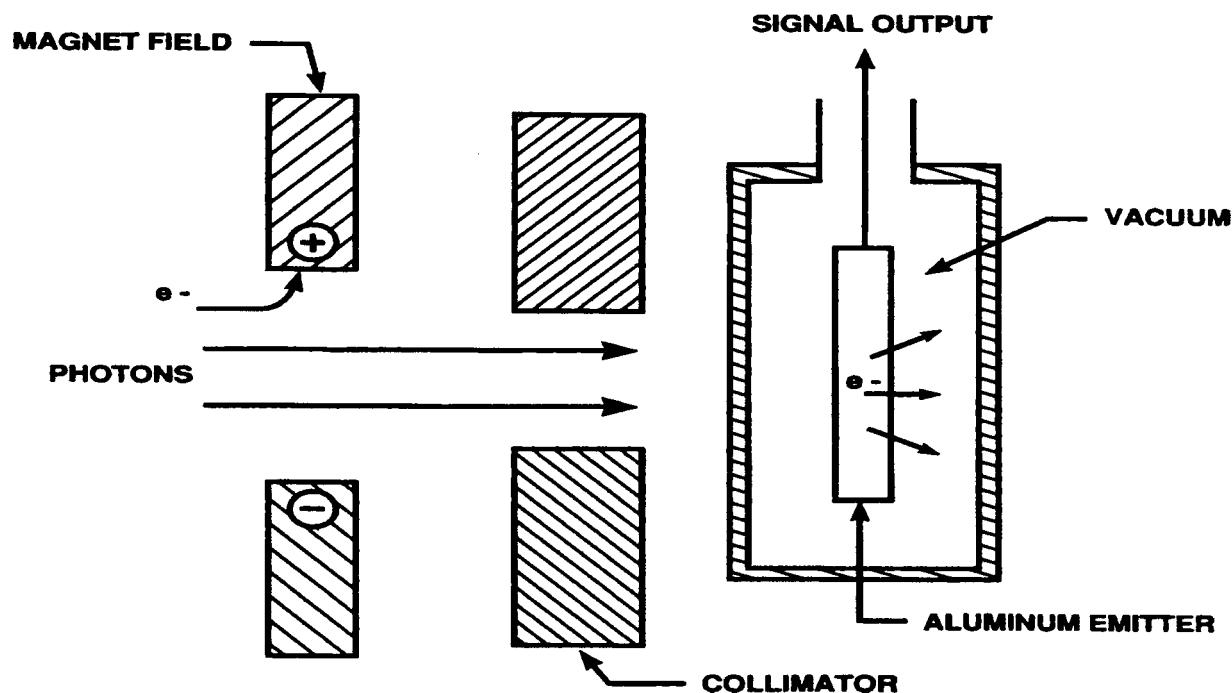


FIG. 5 Vacuum Compton Diode Schematic. Photons interact with the aluminum emitter, creating secondary electrons (e^-) which escape, producing an output signal (+) proportional to the rate of energy deposition.

and are thus not generally particularly well-suited for measurements with low-voltage flash X-ray sources. Appropriately-designed CDs have a response which is relatively constant for incident photon radiation from ~ 0.5 to 5 MeV, the spectral range where the Compton effect dominates. Since Compton electron generation dominates over the broadest range for low-Z materials, dielectric-filled CDs may be less spectrally-

dependent than vacuum CDs. To minimize spectral sensitivity, the dielectric should be at least one electron range thick for the secondary electrons produced by the maximum energy photon, but thin enough to minimally attenuate the photon beam for the lowest energy photon. These conflicting objectives are met reasonably well with a dielectric thickness of 2–3 cm. The collector of a dielectric-filled CD should be fully-stopping for

the maximum energy incident photon, to prevent electron emission from the rear of the collector and consequent signal reduction. Tungsten alloy collectors 3–4 cm thick are adequate to reduce photon transmission to a negligible level.

8.7.4.1 Time Response—The frequency response of a Compton diode is determined by its physical size and electrical characteristics (40). The electron transit time through the detector is defined by the physical dimensions of the collector (emitter), and is typically $\sim 0.1\text{--}0.3$ ns. This electron transit time couples with the impedance characteristics of the CD and signal cable to produce the response time of the detector system. If the impedance of the CD is well-matched to the signal cable (typically $50\ \Omega$), then the detector should have a sub-nanosecond time response. Both dielectric-filled and vacuum CDs have been designed which demonstrate frequency responses > 1 GHz when tested on a 50-ps pulse LINAC source (40–41). For mismatched systems, signal reflections at the cable connector may limit the detector risetime to several ns or longer.

8.7.5 Sensitivity—Typically, sensitivities of $1\text{--}2 \times 10^{-12}$ C/Gy-cm² ($1\text{--}2 \times 10^{-12}$ C/rad-cm²) are observed for both vacuum and dielectric CDs.

8.7.6 Calibration—Compton diodes are generally calibrated using Co⁶⁰ or Cs¹³⁷ sources, in well-defined, carefully collimated geometries. These geometries must be maintained in use to retain the calibration.

8.7.7 Reproducibility—Typical variations in response are 10 % (40, 41).

8.8 Scintillator-Photodetectors:

8.8.1 Introduction—Scintillator-photodetectors are useful as dose rate or pulse shape detectors for flash X-ray sources because 1) they can be made in a large range of sensitivities (10^{-10} – 10^{-5} C/Gy, or 10^{-12} – 10^{-7} A/rad/s) they have a large dynamic range (1,000:1), and 3) their large output signal ($V_{\text{sig}} \leq 800$ V) permits their use in the electromagnetically noisy environment which is often characteristic of large flash X-ray facilities. The useful dose rate range of scintillator-photodetectors is 10^4 – 10^9 Gy/s (10^6 – 10^{11} rad/s).

8.8.2 Principles—The detector consists of a scintillator which is optically coupled to a photodetector. Ionizing radiation is absorbed by the scintillator which emits a pulse of light of an intensity proportional to the absorbed dose rate (43–46). The light is converted to an electrical current pulse by the photodetector which is optically coupled to the scintillator. There are a variety of useful scintillator materials; however, for flash X-ray dosimetry the most frequently used type is the plastic scintillator.

8.8.2.1 A typical plastic scintillator consists of a solid transparent matrix containing a scintillating material. Since the wavelength of the light from most useful scintillator materials often ranges from short wavelength visible light to near ultraviolet, a wavelength shifter which absorbs the light from the scintillator and reemits it at a wavelength which is better matched to the photodetector response is also frequently included.

8.8.2.2 A common photodetector is the planar vacuum photodiode which consists of a photocathode (Type S-4 or S-20) and a fine collecting mesh enclosed in a glass vacuum

envelope. The photodiode collector is biased at a high positive DC potential relative to the photocathode so that all electrons which are photoemitted from the cathode are collected by the anode mesh. As long as the photodiode is operated in the linear region ($I_{\text{lin}} - I_{\text{sat}}$), the output current of the photodiode is proportional to the light intensity, and hence to the absorbed dose rate.

8.8.3 Applications—The detector consists of a plastic scintillator optically coupled to a vacuum photodiode. Both components are contained in a light-tight housing which is provided with signal and bias voltage connectors, and which may also contain a low-inductance biasing network which stores sufficient charge to supply the diode current for the entire duration of the radiation pulse (Fig. 6). In addition, a high-voltage dc power supply and a fast waveform recorder such as an oscilloscope or a transient digitizer are required.

Scintillator:	Pilot-B, NE-102
Photodiode:	ITT FW-114
H.V. =	+ 5 kV
C =	20 nF, 7.5 kV
R =	100 k Ω , 1 W

8.8.3.1 The primary area of applicability of scintillator-photodetectors is as dose rate or pulse shape detectors for high energy flash X-ray sources ($h\nu \geq 300$ keV).

8.8.4 Advantages—Scintillator-Photodetectors should be considered whenever a dose rate detector whose sensitivity lies between that of PIN diodes and Compton diodes is required. Because these detectors produce an output signal of several hundred volts, they are quite immune to electromagnetically coupled noise which is a common problem at flash X-ray facilities.

8.8.5 Limitations:

8.8.5.1 Energy Response—The active photon absorber in this detector is plastic, typically polystyrene or polyvinyl toluene. The mass energy absorption coefficient of most plastics is similar to that of air or water. If a plastic scintillator is used to measure absorbed dose rate in silicon, the photon energy spectrum of the source must not contain any appreciable energy components lower than 300 keV. If there is appreciable photon energy below 300 keV the detector will underrespond to the low energy portion of the source spectrum and will not faithfully reproduce the true pulse shape.

8.8.5.2 Time Response—The time response of the scintillator-photodetector is affected both by the luminescence decay constant of the scintillator and by the time response of the photodiode. The luminescence decay time constant τ , (the light output decays as $e^{-t/\tau}$) is typically about 2 ns for the most common fast plastic scintillators. The time constant of a typical vacuum photodiode in a 50- Ω low-capacitance holder, and biased at 5 kV, is about 0.3 ns. Thus the time response of the detector is essentially the time response of the scintillator.

8.8.5.3 Linearity—The linearity of the detector is the combined linearity of the scintillator and the photodetector. The linearity of the light output of the scintillator is affected both by saturation of luminescence centers and by radiation induced transient darkening of the plastic, and is thus thickness dependent. The light output of a thin plastic scintillator can be linear with absorbed dose rate up to about 10^9 Gy(Si)/s (10^{11}

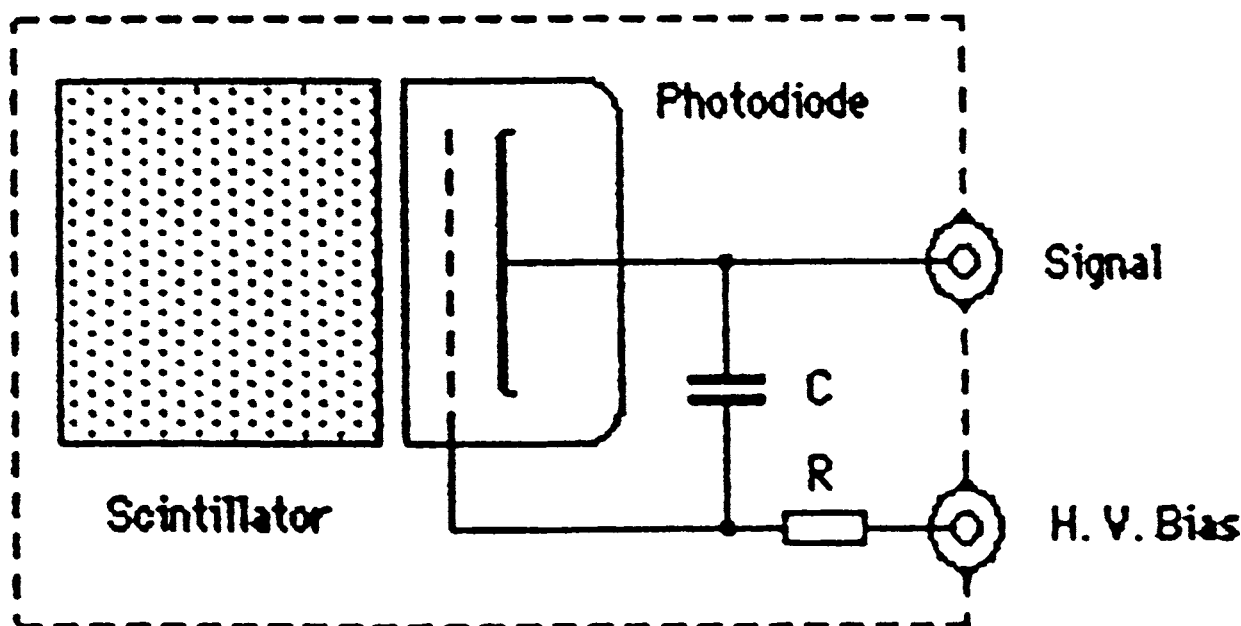


FIG. 6 Diagram of a Typical Scintillator-Photodetector.

rad(Si)/s). The diode current of a vacuum photodiode is linear up to $I_{\text{sat}}/2$. I_{sat} depends on the electrode spacing, photocathode area, and bias voltage. For a typical vacuum photodiode biased at 5 kV, I_{sat} is about 32 A; this detector would therefore operate linearly up to a diode current of about 16 A. Since a planar vacuum photodiode closely resembles an ideal space-charge limited diode, the saturation current is related to the photodiode parameters and the bias voltage by the Langmuir-Child relationship:

$$I_{\text{sat}} = 2.33 \times 10^{-6} (AV^{3/2})/d^2 \quad (12)$$

where:

A = is the area of photocathode, m^2 ,
d = is the cathode-collector spacing, m, and
V = is the bias voltage, V.

8.8.6 Sensitivity—The sensitivity of scintillator-photodetectors spans the range of 10^{-10} – 10^{-5} C/Gy (10^{-12} – 10^{-7} A/rad/s). The useful dose rate range of scintillator-photodetectors is 10^4 – 10^9 Gy/s (10^6 – 10^{11} rad/s). For maximum pulse shape fidelity, the photon spectrum of the flash X-ray generator must be substantially above 300 keV.

8.8.7 Calibration—Scintillator-photodetectors are normally not used as calibrated dose-rate detectors. They are usually used as a relative indicator of pulse shape. This information together with the absorbed dose (as measured by an integrating dosimeter) is sufficient to calculate the dose rate as a function of time.

8.8.8 Reproducibility—Typical uncertainties are 4 % (44-46)

8.9 Photo-Conductive Detector Dosimeters:

8.9.1 Introduction—A photo-conductive detector (PCD) can be thought of as a resistor, whose resistance is inversely proportional to the deposited dose rate (47-50). PCDs are often made from diamond or GaAs, but other large bandgap semi-

conductors are possible (51, 52). Once the detector has been procured or manufactured, the system is relatively straightforward to implement in its simplest form requiring only materials available at most flash X-ray facilities. Indeed the circuit and hardware used are identical to that of the PIN diode (see Fig. 3) although PCDs are often biased at higher voltages. The primary drawbacks to PCDs are that dose must be converted from dose in carbon or GaAs to dose in silicon and the requirement for frequent calibration due to drift or charging of the insulating materials. PCDs are a very common dose rate sensor at low energy X-ray facilities, such as plasma radiation sources, but they also are becoming common at flash X-ray facilities, although they must be packaged differently.

8.9.2 Physical Principles—A useful PCD is a large bandgap semiconductor with a short lifetime for the electrons in the conduction band. The large band gap limits the room temperature conductivity and the short life time ensures that the number of electrons in the conduction band is proportional to the generation rate. Typical resistances when not in a radiation field are $> 1\text{G}\Omega$ and lifetimes are < 500 ps (53). The risetime of the PCD is on the order of the lifetime of the carriers. With these properties in mind, the measured resistance of the PCD is inversely proportional to the dose rate and the PCD will obey Ohm's Law. Hence, unlike a PIN diode, there will be virtually no leakage current and for a given dose rate the magnitude of the signal is proportional to the applied voltage. Unfortunately, unlike the PIN diode, this does mean that a change in the voltage will result in a change in the measured response.

8.9.2.1 PCDs tend to be small, yet macroscopic in size with each dimension being ~ 0.1 – 3 mm. The sensitivity and speed of the PCD can be varied by neutron damage of the material to further shorten the lifetime. Because of their lower sensitivity, PCDs often are used at high dose rate facilities. However, the

high dose rate can cause inferences. Because the resistivity of the PCD is often higher than the air, the air conductivity will mask the response of the PCD. Therefore, the PCD is often potted in a low radiation response dielectric to preclude air conduction effects. However, this often results in another interference, that is, trapped charge in the dielectric. This can affect the calibration of the PCD and therefore when such potting is used the PCD must be irradiated several times at a particular voltage polarity and magnitude before a constant response and calibration value is obtained. Changing this value will result in further calibration.

8.9.2.2 Because the calibration can slowly drift with time due to changes in trapped charge or annealing of displacement damage in the material, they must be calibrated frequently, ~monthly. In this case, they are typically calibrated by comparison to a better-controlled dosimeter, such as a TLD. Once calibrated, the PCD may be used repeatedly in the same radiation field.

8.9.3 *Applications*—PCDs provide the capability to obtain dosimetry information on-line, without entry to the flash X-ray cell or opening up a large experimental configuration. When located at fixed distances from a flash X-ray machine, these detectors perform well as dose-rate and timing monitors. PCDs may also be employed as a flash X-ray output diagnostic or for field mapping. Because they are often less sensitive than the PIN diode, they are often used in the region of maximum beam intensity.

8.9.3.1 Correct electronic configuration of the PCD is essential for the proper functioning of the system. Fig. 3 shows a typical setup with the PCD substituted for the diode. The biasing capacitor may be located close to the PCD to maintain biasing voltage during the flash X-ray pulse. Sometimes two capacitors are used: a larger to supply the bulk of the charge, and a smaller, low-inductance capacitor for rapid response. The bias voltage supply is decoupled from the circuit during the pulse by a resistor. Since many of the components of the PCD system are exposed to at least part of the flash X-ray radiation environment, their response to the radiation may form part of the system response. A useful technique to help separate the PCD response from noise is to expose the system to flash X-ray pulses with all components in place except for the PCD. The PCD is replaced either with an open circuit (such as using an epoxy insulator) or with a short circuit. With the open circuit, the system may be run under bias. Signals obtained with this setup are subject to noise, and must be maintained small compared to the signal obtained with the PCD in place. Filling the electrical connectors, exposed to the radiation, with vacuum grease to reduce the air conductivity in the cable plant can reduce extraneous responses.

8.9.4 *Advantages*—PCDs have the advantages of small size, moderate cost, and medium to low sensitivity. In some cases, the PCD material may be selected to match the material of interest, such as GaAs. However, one must be extremely careful that the material surrounding the diode does not interfere with the desired response and that electron equilibrium is obtained.

8.9.5 *Limitations*—Even though PCDs are often less sensitive than PIN diodes, they can still debias dose rates typical of

flash X-ray testing. This debiasing of the voltage occurs when the voltage at the data recorder is a significant fraction of the applied bias. When this happens the bias across the PCD decreases. Therefore, the measured signal must be kept small (~5–10 %) compared to the bias voltage or corrected using the following formula:

$$V_{corr} = V_{meas} / (1 - V_{meas} / V_{bias}) \quad (13)$$

where:

V_{corr} = the corrected voltage after removing the debias effect,

V_{meas} = the voltage measure at the recording instrument, and

V_{bias} = the bias voltage, V.

8.9.5.1 In some cases, for example diamond, the PCD material will not be the same as the DUT. Hence, it may be necessary to know the radiation spectrum and use a radiation transport code to infer the equilibrium dose or the dose in the DUT.

8.9.5.2 The sensitivity of the PCD can change if exposed to higher temperatures, especially if the PCD has been previously been neutron damaged. Care must be taken to keep the PCD at or below room temperature. Also because of charging effects in the surrounding insulator care must be taken to calibrate and use the PCD with a consistency voltage both in polarity and magnitude.

8.9.6 *Sensitivity*—PCDs are available for measuring dose rates from below 2×10^5 up to 10^{10} Gy(Si)/s (2×10^7 up to 10^{12} Rad(Si)/s) with rise times < 1 ns. The sensitivity of the PCD is determined by its material and its geometry. The resistivity of the material can be increased with neutron damage, if desired. The sensitivity can also be adjusted by the appropriate choice of geometry using the following formula:

$$R = \rho L / A \quad (14)$$

where:

R = the resistance,

ρ = the resistivity of the material,

L = the distance between the voltage contacts, and

A = the cross sectional area.

By appropriate choice of L and A a large range of sensitivities can be obtained for a given material.

8.9.7 *Calibration*—The calibration of the PCD may best be performed by observing its behavior in a beam of known intensity and time variation. The intensity can be lower than expected in the flash X-ray environment, and the time variation faster. A LINAC with microstructure is not recommended because the PCD, due to its short carrier lifetime, can respond to the microstructure and unless the instrumentation can record it properly will give erroneous results.

8.9.7.1 Alternatively, the PCD response may be integrated over time to find the total delivered dose. This dose is then compared to a total-dose detector (such as a TLD) exposed simultaneously with the diode. Note that in this case the calibration is in dose TLD and the calibration is only good for spectra sufficiently similar, as determined by radiation transport calculations, to the calibration source.

8.9.8 *Reproducibility*—Operating in the linear range, PCDs can measure dose rates with a precision of better than 5 %. The

main limiting factor is electrical noise introduced in the cabling. Because the calibration can drift over time, PCDs should be recalibrated before use in any new test series or if there has been loss of control of the detector (unknown temperature or bias conditions).

8.9.8.1 Biases in the result will be introduced by the calibration procedure. The reference dosimeter used to calibrate the diode may have its own bias. Measurements to within 10 % of true values may be obtained.

9. Minimum Documentation Requirements

9.1 Record or reference the flash X-ray source characteristics such as a maximum photon energy and spectral shape, peak fluence rate, fluence rate pulse shape and time dependence, and beam uniformity and profile.

9.2 Record or reference the dosimeter manufacturer, type, batch number, and instruments used for analysis.

9.3 Record dosimeter calibration data, including date, a description of the calibration facility, and procedure.

9.4 Record the irradiation environmental conditions for the dosimeters, including temperature, pressure (if other than atmospheric), relative humidity, and surrounding atmosphere (if other than air).

9.5 Record or reference the method used to convert the dosimetry measurements to absorbed dose values in the material of interest.

9.6 Record of the value and the assigned uncertainty of the absorbed dose measurement for each irradiation. Identify systematic and random components of the uncertainty.

NOTE 2—The factors listed in 9.2-9.6 may not all be applicable to some of the active type dosimeters. For example, it is generally not necessary to check the calibration for calorimeters. Therefore, only the factors that are appropriate for the specific dosimetry system being utilized should be recorded or referenced.

10. Precision and Bias

10.1 To be a meaningful, a measurement of absorbed dose shall be accompanied by an estimate of the uncertainty in the measured value. Factors contributing to this uncertainty may be separated into two types, precision (random) and bias (systematic). The uncertainty should be stated as an estimate of the probable limits of error, combining both precision and bias.

NOTE 3—Measurement uncertainty is described by a precision and bias statement in this standard. Another acceptable approach is to use Type A and B uncertainty components (54, 55). This Type A/B uncertainty specification is now used in International Organization for Standardization (ISO) standards and this approach can be expected to play a more

prominent role in future uncertainty analyses.

10.2 Precision

10.2.1 For a measurement of absorbed dose, the precision (random uncertainty) of the measurement process is due to at least two effects. These are variations in intrinsic response of the dosimeters (or dosimeter) and of the associated instrumentation. The precision due to these two effects may be determined by replicate measurements of a specific absorbed dose. The number of replicates is determined by the variability in dosimetry system response and the specified confidence level (see Practice E 668 for general guidance on determining this number). There may be other sources of random uncertainty contributing to the total estimate of the precision of the measurements. Assuming normal distribution, all values of such components of the uncertainty may be summed in quadrature (that is, the square root of the sum of the squares) and the results reported at the specified confidence level. Determine the precision for the dosimetric system at the time of calibration (or other appropriate time) and repeat the procedure if a change takes place that might affect the precision.

10.3 Bias

10.3.1 All non-random factors contributing to the total uncertainty of a measurement are called bias (or systematic uncertainty). These will include the bias (with respect to the true value) associated with the calibration facility as reflected by the reference standard dosimeter used to establish traceability to the national standard. It will also include the effect of any differences between calibration and test conditions.

10.3.2 Identify sources of bias and make corrections for those of known magnitude. Estimate the remaining biases. Usually it is not possible to determine the actual probability distribution of the remaining biases. One approach is to assume an approximately normal distribution for such biases and treat them as though they are random in nature. Then, sum the remaining biases in quadrature (this sum corresponding roughly to a standard deviation) and report the results at the same confidence level as that for the precision.

10.4 Precision and bias may be combined in quadrature as a statement of total uncertainty at a specific confidence level.

11. Keywords

11.1 calorimeter; Compton diode dosimetry; dosimetry; optical waveguide dosimetry; photoconductive detector (PCD); PIN diode dosimetry; pulsed X-ray source; radiochromic dosimetry system; scintillator-photodetector dosimetry; thermoluminescence dosimetry



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