Determination of Photon Conversion Factors Relating Exposure and Dose for Several Extremity Phantom Designs

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DETERMINATION OF PHOTON CONVERSION
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FOREWORD

Accurate, reliable estimates of absorbed dose are essential to the management of each individual radiation worker's exposure. Over the past few years, accreditation and application research programs focused on the accuracy of dosimeters used for external, whole-body measurements with little attention to dosimeters used for extremity measurements. Anticipated changes in basic radiation protection standards are mandating that greater emphasis be placed on extremity measurements. Specifically, promulgation of the proposed revision to the Environmental Protection Agency's "Radiation Protection Guidance to Federal Agencies for Occupational Exposures" would lower the exposure limit for extremities from 75 to 50 rem, and would require an improved ability to differentiate between penetrating and nonpenetrating exposures. Additionally, the International Commission on Radiological Protection has recommended that the quality factor for neutrons be raised from 10 to 20, which will then effectively reduce the allowable neutron exposure to the extremities. In view of these developments, the Health Physics Group, Office of Nuclear Safety, is accelerating its efforts in extremity dosimetry development.

In 1982, the Health Physics Society Standards Committee (HPSSC) initiated the draft ANSI P/N13.32, "Standard for the Performance Testing of Extremity Dosimeters." The standard, in part, is expected to consider the results of a study supported by the Department of Energy (DOE) that was intended to provide information on the selection of an appropriate calibration phantom for extremity dosimeters and the need to include exposure-to-dose conversion factors (C_x) for extremity dosimetry. When the HPSSC has completed its effort, DOE plans to evaluate the standard for incorporation into the DOE Laboratory Accreditation Program for Personnel Dosimetry.

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EXECUTIVE SUMMARY

This report presents the results of measurements of dosimetric properties of simple extremity phantoms suitable for use in extremity dosimeter performance testing. Two sizes of phantoms were used in this study. One size represented the forearm or lower leg and the other size represented the finger or toe. For both phantom sizes, measurements were performed on solid plastic phantoms and on phantoms containing simulated bone material to determine the effect of backscattered radiations from the bone on the surface dose.

Exposure-to-dose conversion factors (C_x factors) were determined for photon energies ranging from 16 to 1250 keV (average for ^{60}Co). The effect of the presence of a phantom was also measured for a $^{90}\text{Sr}/^{90}\text{Y}$ source. The ratio of the dose to thin thermoluminescent dosimeters on the surface of the phantoms to the dose to the dosimeters in air was determined using replicate measurements. The measured ratios were multiplied by well-known values to arrive at C_x factors.

Significant differences in the measured C_x factors were found among the phantoms investigated. The factors for the finger-sized phantoms were uniformly less than for the arm-sized phantoms. The factors for the finger phantoms did not vary significantly between phantoms that had simulated bone inserts and those that did not. The most likely cause of this insensitivity to construction for finger phantoms was the small size of the phantom. Only a small phantom backscatter contribution was made to the surface dose (less than 20% for all energies) and minor variations in this contribution are relatively unimportant. The C_x factors for arm phantoms with either bone-equivalent or aluminum inserts were significantly less than factors for solid plastic phantoms over the energy range of 40 to 100 keV. This effect was attributed to the preferential absorption of the lower energy backscattered photons in the higher atomic number material of the insert.

Because of the small effect of simulated bone inserts on the C_x factor for finger phantoms, we recommend a solid plastic phantom for use in calibration. Due to the relatively large effect of simulated bone inserts at the middle x-ray energies for the arm phantom, a phantom with an insert of higher

atomic number material is recommended. This should result in a closer correspondence between surface dose on the extremity phantom and the extremity of the worker for the same radiation field conditions. The position of the bone-simulating material below the surface of the phantom is more important than its size or its level of bone equivalency. For simplicity, we recommend that a phantom with an aluminum insert be used. The diameter of the insert may vary between 48 and 60 mm without greatly affecting the C_x factors.

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1.0 INTRODUCTION

The response of personnel dosimeters can vary widely in the absence of guidance specifications for their calibration and energy response determinations. The universal use of specified and easily-constructed calibration geometries will provide a common means for comparing dosimetry results and striving for improvements. No such guidance exists for specifying calibration geometries and dose for the calibration of extremity dosimeters.

This report describes prototype extremity phantoms and their dose specification factors (C_x factors) used in a study to aid the Health Physics Society Standards Committee to select phantom designs and C_x factors for the development of a draft American National Standards Institute (ANSI) standard entitled "Standard for the Performance Testing of Extremity Dosimeters (ANSI P/N13.32)." This study also adds to the available information on the calibration of extremity dosimeters for use by the Department of Energy (DOE).

The approach to defining standard calibration specifications is parallel to that used for the performance testing of personnel body dosimeters in the ANSI Standard N13.11, "Criteria for Testing Personnel Dosimetry Performance" (ANSI 1983). That standard specifies the phantom types; the irradiation geometries; and photon, neutron, and beta energies for the conduct of performance tests.

Even though the calibrations of dosimeters for occupational environments very often differ from those used for the laboratory test, the test calibration represents a reference point for all dosimeters. As refinements in personnel dosimetry are made, a common goal will be the improvement of the dose specification for the test irradiations and the improvement of the designs of the personnel dosimeters. These improvements will minimize the difference between the calibrations for the occupational environments and the performance tests. To better achieve this goal, the test specifications should be made as realistic as practicable.

Dose measurements were performed to determine C_x factors for a variety of simple extremity phantoms. The phantoms were chosen to represent the forearm or the lower leg, or a finger or toe. The phantoms were constructed of solid

plastic (polymethyl methacrylate, PMMA) or of a PMMA cylindrical shell with a core of higher atomic number material. Measurements were performed with photon source energies ranging from 16 keV to 1.25 MeV (average) and a $^{90}\text{Sr}/^{90}\text{Y}$ beta source. The measurements will be used to provide approximate C_x factors for use in calibrating extremity dosimeters on simple phantoms, to estimate the significance of C_x factors for extremity dosimetry, and to help the Health Physics Society Extremity Standards Committee choose phantom designs.

2.D EXPERIMENTAL DESIGN AND METHODS

Dose at the surface of extremity phantoms was measured for many source energies. The approach used to measure dose and to derive the C_x factors is described in Section 2.1. The phantom design and construction is described in Section 2.2. The use of the dosimeters is described in Section 2.3 and the experiment design is described in Section 2.4.

2.1 APPROACH

The measurement of surface dose for an extremity phantom is not easily performed with an instrument. The sensitive volume must be very small to avoid perturbing the radiation field at the measurement location. Rather than expend a considerable effort performing a measurement using an instrument with a tissue-equivalent response, a simple approach using thermoluminescent (TL) dosimeters was chosen. An instrument measurement may have resulted in smaller systematic errors, depending on the level of tissue equivalency that could have been achieved. The precision of the measurements, however, would probably have been worse than was obtained by using the TL dosimeters. The potential problem of systematic errors affecting the TL dosimeter results were minimized by performing repetitive measurements on the extremity phantoms and performing similar measurements on the body phantoms, for which tabulated C_x factors are available. The measurement uncertainty analysis is presented in Section 3.3.

The TL dosimeters were selected based on availability, thickness of the sensitive layer, and dosimetric properties. The available TL dosimeter with the thinnest effective depth (quoted as 5 mg/cm^2 of LiF) was manufactured by Vinten Instruments Limited (Dutt et al. 1984). The Vinten dosimeters were tested for dosimetric properties and reproducibility, resulting in a technique to successfully reuse these dosimeters, even though they are designed to be discarded after one use. This procedure is described in Section 2.3 and the appendix. The nominal 5-mg/cm^2 dosimeter was wrapped in a 5-mg/cm^2 plastic covering to achieve measurements averaged over a 5- to 10-mg/cm^2 depth.

Dose to tissue cannot be directly measured by LiF TL dosimeters because their response is not directly proportional to tissue dose as a function of source type and energy. The measurements were not directly interpreted as dose. The TL dosimeter data were used to determine the change in dose with and without the phantom in place. The ratios of dose to a small amount of tissue per dose to air are well-known tabulated quantities for the photon sources of interest. Dose to air and exposure are directly related by a unit conversion. These ratios were used to calculate the extremity phantom C_x factors.

Phantom design was motivated by the desire to establish a simple design as the standard for future reference calibrations. The design should be easily specified, manufactured, and used. The relationship between the simple designs used for reference calibrations and more realistic phantoms designed to represent occupational conditions should be determined by special measurement.

2.2 PHANTOM DESIGN

The phantoms that were considered candidates for use in an extremity standard were chosen to be radially and axially symmetric (right circular cylinders or axially colinear, nested, right-circular cylinders). Other asymmetric phantoms were chosen to investigate the effects of the presence of bone near the surface and to investigate the dose to the ends of the finger tips. All phantoms, except the finger tip phantom, were mounted at right angles to the beam direction and extended beyond the beam edges. This eliminated potential effects of phantom-to-air discontinuities on the phantom dose arising from the finite extent of the phantom. The advantage of this technique is to make phantom design and mounting less critical. For the uncollimated ^{137}Cs exposures, a length of 600 mm was chosen to approximate a phantom of infinite extent. The exception was the finger phantom designed to determine dose to the tip of the finger. This phantom was made short with a rounded tip on one end. Exposures were performed with the finger tip pointing toward the source.

Polymethyl methacrylate (PMMA) was chosen as the construction material for the plastic portion of the extremity phantoms. This material is specified

for the simulated whole-body phantoms in the ANSI N13.11 Standard (ANSI 1983) and is readily available under several tradenames, such as Plexiglas® and Lucite®.

For those phantoms used to study the effects of bone, two materials were chosen to represent the higher atomic-number composition of bone. The first material was a bone-equivalent plastic mixture, mix code SB5, described by White, Martin and Darlison (1977). The elemental composition and specific gravity are given in Table 2.1. This mixture was designed to have a density and atomic number close to that of bone. It was assumed for this study that the simulation adequately represented the true effects of bone.

The second material was studied to investigate the possibility of using a readily available material to simulate the effects of bone for standard phantom specifications. The material chosen for study was aluminum. Aluminum is readily obtainable with negligible variations in composition, is easily machined, and has an atomic number greater than the average atomic number of tissue. By comparing solid plastic, plastic with a bone-equivalent center, and plastic with an aluminum center, estimates of the true dose to the phantom surface and the sensitivity of the C_x factors to phantom construction were obtained.

The properties of bone, aluminum, and muscle are compared in Table 2.2. The photoelectric effect is important for photons below about 50 keV and Compton scattering is important between about 50 keV and 10 MeV. The photoelectric mass attenuation coefficient is approximately proportional to NZ^3 , where N is the electronic density in electrons/gram and Z is the effective atomic number (White 1978). The linear attenuation (equal to the mass attenuation times the density) is the quantity used to compare the Compton reaction rate in bone, aluminum, and muscle. The linear attenuation coefficient due to Compton scattering is directly proportional to the number of electrons/cm³ of material. Table 2.2 lists the relevant properties of bone, muscle, and aluminum, including the ratio of the photoelectric mass attenuation of muscle and aluminum to bone, and the ratio of the Compton linear attenuation

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TABLE 2.1. Composition of Bone-Equivalent Plastic Mix SB5
(White, Martin and Darlison 1977)

<u>Element</u>	<u>Percent Weight</u>
H	2.6
C	30.6
N	1.0
O	38.9
Cl	0.1
Ca	26.8
Specific Gravity = 1.87	

TABLE 2.2. Comparison of the Properties of Bone,
Aluminum, and Muscle

<u>Quantity</u>	<u>Bone</u>	<u>Aluminum</u>	<u>Muscle</u>
Density (g/cm ³)	1.8	2.7	1.04
Electronic Density (electrons/cm ³)	3.21×10^{24}	2.9×10^{24}	3.31×10^{24}
Ratio of Photoelectric Mass Attenuation to Bone	1.0	1.6	0.35
Ratio to Compton Linear Attenuation to Bone	1.0	1.36	0.60

coefficient of muscle and aluminum to bone. From Table 2.2, we see that the aluminum insert will produce a more pronounced effect than bone. However, the magnitude of effects for the phantom with the bone insert will be between the values for the solid phantom and the phantom with the aluminum insert.

Little helpful guidance in choosing dimensions for extremity phantoms exists. The Report of the Task Group on Reference Man (ICRP 1975) lists fractional surface areas for the extremities and the total body. Assuming cylindrical shapes for the extremities, the derived diameters are 80 mm for the arm and 110 mm for the leg. These numbers are the average over the total extremity, however, and are an overestimate for the lower extremities. As a check, extremity diameters were also determined for a sampling of individuals at PNL. Average diameters for the wrist, mid-forearm, elbow, and ankle were

58, 82, 91, and 78 mm, respectively. The sample standard deviation of the diameter for the forearm was calculated to be 10 mm. The 73-mm diameter for the arm/leg phantom was chosen as a compromise between the ankle, wrist, and forearm diameters.

Because of the small mass of the finger phantom, the exact diameter is not critical. Furthermore, for workers, the presence of the other fingers and the hand will perturb the radiation field more than differences in finger diameter. The 19-mm diameter for the finger phantom was chosen because of availability (3/4-in. stock) and because this diameter could be used easily with typical ring dosimeter designs.

The dimensions and phantom types are listed in Table 2.3. Phantoms with "A" codes were designed to represent the forearm or the lower leg. Phantoms with "F" codes were designed to represent fingers or toes. Phantom F4 simulated a finger tip. All "A" or "F" phantoms had the same outer diameter and consisted of PMMA except for the specified inserted material.

2.3 TL DOSIMETER HANDLING

The use of the thin Vinten dosimeters required the development of special handling techniques. The dosimeters as supplied by the manufacturer were intended for one readout only. Special techniques were formulated to overcome the following problems: 1) the dosimeters could not be annealed in the plastic encasement provided by the manufacturer; 2) the dose response between dosimeters varied by as much as a factor of three; 3) the dosimeter off-gassing during readout deposited a film on the reader optics causing a steadily changing reader calibration; and 4) dosimeter curling during the readout process caused incomplete readout.

Annealing and reuse procedures were developed to allow the use of sensitivity calibration factors. The original plastic packaging was removed, leaving the LiF phosphor embedded in a high-temperature-resistant adhesive. The phosphor was then annealed at high temperature and handled similar to other LiF dosimeter elements. A special dosimeter-retaining clip was used to minimize dosimeter curling and a 5-mg/cm² thick black polyethylene cover was added. The manufacturer's estimated thickness of 5 mg/cm² for the TL element implies net monitoring depths of 5 to 10 mg/cm² for this study. The

TABLE 2.3. Extremity Phantom Designs

Phantom Code	Description
A1	Solid PMMA with 73-mm (2-7/8-in.) diameter
A2	PMMA tube with 73-mm outer diameter Aluminum insert with 60-mm (2-3/8-in.) diameter
A3	PMMA tube with 73-mm outer diameter Bone-equivalent insert with 60-mm diameter
A4	PMMA tube with 73-mm outer diameter Aluminum insert with 48-mm (1-7/8-in.) diameter
A5	PMMA tube with 73 mm outer diameter Bone-equivalent insert with 48-mm diameter
A6	PMMA cylinder with 73-mm outer diameter and 25-mm (1-in.) offset hole, 6 mm (1/4 in.) beneath surface Bone-equivalent insert with 25-mm diameter
F1	Solid PMMA with 19-mm (3/4-in.) diameter
F2	PMMA tube with 19-mm outer diameter Aluminum insert with 13-mm (1/2-in.) diameter
F3	PMMA tube with 19-mm outer diameter Bone-equivalent insert with 13-mm (1/2-in.) diameter
F4	PMMA tube with 19-mm outer diameter and domed top to represent a finger tip Bone-equivalent material molded into cavity (13-mm diameter)

dosimeters were reused up to seven times before they were discarded. Pre- and postexperiment calibration factors were developed to determine long-term sensitivity variations in the reader calibration. Details of the procedure are given in the appendix.

The effective thickness of the Vinten dosimeters was checked using their response to beta sources. The measured ratio of the response to ^{204}Tl and $^{90}\text{Sr}/^{90}\text{Y}$ was 0.96 ± 0.11 . The estimated effective width is 0 to 20 mg/cm^2 , based on the measured depth dose curves for those sources. The variation in effective thickness for the sampled dosimeters contributed to the uncertainty of the measurement. The verified effective width is sufficient to meet the goals of the dose-response measurements for photons.

2.4 EXPERIMENTAL DESIGN

The TL dosimeters were exposed, read out, and annealed in sets of 60 to 80 dosimeters. Both in-air and on-phantom exposures were included with each set, as well as 10 dosimeters exposed to ^{137}Cs under 100 mm of plastic and 2 unexposed dosimeters. A total of 5 exposure sets were used, each for up to 7 cycles. Background signals ranged from 1% to 2% of the exposed dosimeter readings. The ^{137}Cs -calibrated dosimeters were used to relate measurements made in separate cycles and to track dosimeter degradation. All dosimeters for each set were exposed to ^{137}Cs under 100 mm of plastic at least twice, from which TL element sensitivity factors were developed. The on-phantom-to-in-air ratios were calculated following corrections for TL element sensitivity and background subtraction.

Not all combinations of beam type and energy and phantom construction were used for the experiment because of limited time and resources. The data for the solid PMMA phantoms (A1 and F1) were collected for nearly all energies to provide reference energy response curves. Data for all phantoms were collected for three x-ray techniques (16, 60, and 120 keV) to determine the effects of phantom construction in the energy ranges most likely to result in observable differences. Measurements at additional energies were performed for the phantom with the 60-mm bone-equivalent insert (A3) to better determine the energy dependence of observed differences compared to results for the solid PMMA arm phantom. The dependence of the technique on variations in beam setup and TL dosimeter readout were tested by doing many replicate measurements for the same setups. Differences because of setup variations were not observed and were presumed less than the random uncertainties in the TL dosimeter readings.

The irradiation source properties are listed in Table 2.4. The dependence of the response to photon energy was measured using K-fluorescence x rays (Kathren, Rising and Larson 1971; IOS 1979) ranging from 16 to 76 keV, heavily filtered bremsstrahlung techniques with effective energies of 120 and 211 keV (NBS 1981) and ^{137}Cs and ^{60}Co isotopic sources. Measurements were also performed using three of the bremsstrahlung techniques (M30, S60, and M150) specified in ANSI N13.11 (1983). The source designations for the K-fluorescence

TABLE 2.4. Irradiation Source Properties

K-Fluorescence X Rays

<u>Source Designation</u>	<u>Average Energy, keV</u>
K16	16
K23	23
K31	31
K40	41
K49	50
K59	60
K75	76

Bremsstrahlung X Rays

<u>Source Designation</u>	<u>Average or Effective Energy, keV</u>
M30	20
S60	36
M150	70
H150	120
H250	211

Isotopic Sources

<u>Designation</u>	<u>Source Type</u>	<u>Average Energy, keV</u>
^{137}Cs	Photon	662
^{60}Co	Photon	1250
$^{90}\text{Sr}/^{90}\text{Y}$	Beta	800

beams follow the energy of the $K_{\alpha 1}$ x-ray transition and differ from the average energy of the technique. Measurements were also performed with a $^{90}\text{Sr}/^{90}\text{Y}$ beta particle source.

The TL dosimeter readings were used to determine on-phantom to in-air ratios that represented the effects of the presence of the phantom. The value estimated by the TL measurements was the dose to tissue inside the phantom at the 7-mg/cm^2 depth divided by the dose to a small mass of tissue in air. The errors due to the TL response differences compared to tissue dose were assumed to nearly cancel when the ratio was taken. The C_x values were calculated using the following formula:

$$C_x = BF \cdot f \quad (2.1)$$

where BF, or backscatter factor, is the ratio of the tissue dose on phantom to the tissue dose in air and f, or f factor, is the ratio of the tissue dose in air to the exposure in air. The backscatter factor was approximated by the measured on-phantom to in-air ratios. The f factor is a tabulated, slowly varying function of energy. The f factors for muscle tissue were taken from NCRP 69 (NCRP 1981) and are listed in Table 2.5 for the beam energies and types used for this study.

TABLE 2.5. Exposure-to-Dose Conversion Factors
(Small Mass Limit)

<u>Source Designation</u>	<u>Conversion Factor for Muscle (f), rad/R</u>
K16	0.871
K23	0.864
K31	0.863
K40	0.868
K49	0.886
K59	0.901
K75	0.925
M30	0.865
S60	0.866
M150	0.915
H150	0.948
H250	0.957
^{137}Cs	0.958
^{60}Co	0.958

3.0 RESULTS

The measurement results and the uncertainty analysis are presented. The use of TL dosimeters for the determination of C_x factors was checked by performing measurements on slab body phantoms and comparing the results to tabulated factors. These measurements are described in Section 3.1 and are used in the uncertainty analysis presented in Section 3.3. The extremity phantom measurements are described in Section 3.2.

3.1 TECHNIQUE VERIFICATION MEASUREMENTS

The measurement technique used is not exact because of the nontissue-equivalent response of the TL dosimeters. Measurements were performed on slab body phantoms and compared to tabulated C_x values to help determine an upper limit to the systematic errors associated with this method. The systematic measurement errors for the extremity phantom measurements are smaller than for the body phantom measurements because the surface dose due to scattering in the phantom is less.

The energy dependence of the response of the dosimeters was determined at a given photon energy by taking the ratio of the TL readings for the in-air exposures and ^{137}Cs -calibration exposures. Because all expose-read-anneal cycles included the ^{137}Cs calibration, these data represent direct response ratios without further corrections. The data is presented in Table 3.1 along with the uncertainty associated with the variability of the TL readings. The energy response function is plotted in Figure 3.1. The results are normalized to the response for the ^{137}Cs in-air exposure for convenience. The over-response of the TL signal reaches a maximum of approximately 1.6 at low energies (20 to 30 keV). The data deviates from a smooth curve only to an extent consistent with measurement error.

The C_x factors for the body phantoms were determined as described in Section 2. The results are given in Table 3.2. Only shallow C_x factors were determined in correspondence to the extremity measurements. The first column presents the source designation. The K-fluorescence sources were used up to K59 technique along with three of the bremsstrahlung techniques specified in

TABLE 3.1. Energy-Dependent Response of the Thin Dosimeters

Beam	Average or Effective Energy, keV	In-Air Response/Dose to Air, Normalized to ^{137}Cs
K16	16	1.52 ± 0.02
K23	23	1.58 ± 0.09
K31	31	1.50 ± 0.09
K40	40	1.47 ± 0.03
K49	49	1.44 ± 0.04
K59	60	1.32 ± 0.02
K75	77	1.18 ± 0.04
H150	120	1.24 ± 0.02
H250	211	1.18 ± 0.02
^{137}Cs	662	1.00 ± 0.02
^{60}Co	1250	1.01 ± 0.02

FIGURE 3.1. Energy-Dependent Response for the Thin TL Dosimeters.
(Data normalized to the results for ^{137}Cs , 662 keV.)

TABLE 3.2. Comparison of Shallow C_x Factors for Body Phantoms

Source Designation	Slab Phantom, PNL Plastic			Slab Phantom, PMMA	ICRU Sphere
	Monte Carlo ^(a)	TEEC ^(b) Measurement	Present Measurement	Present Measurement	TE ^(c) Material Monte Carlo ^(d)
K16	0.88	1.08	0.85(03) ^(e)	0.93(03)	0.90
K23	1.01	1.07	1.03(02)	1.00(02)	0.99
K31	1.13	1.07	1.08(05)	1.16(04)	1.13
K40	1.33	1.20	1.33(08)	1.39(03)	1.34
K49	1.53	1.36	1.34(03)	1.59(04)	1.48
K59	1.62	1.48	1.42(07)	1.69(05)	1.52
M30	-	1.08	-	0.94(03)	0.92
S60	-	1.15	-	1.32(07)	1.14
M150	-	1.41	-	1.47(06)	1.43

(a) Values taken or interpolated from R. F. Nelson (1984).

(b) Measurements performed with a tissue-equivalent extrapolation chamber (TEEC) (Yoder et al. 1979). Some of the values are interpolated.

(c) TE is tissue equivalent.

(d) Values taken or interpolated from ANSI N13.11 (1983). They are based on Monte Carlo calculations using the ICRU sphere (Dimbylow and Francis 1979).

(e) 0.85(03) is read 0.85 ± 0.03 .

ANSI N13.11 (1983). The next three columns contain values for the slab phantom employing plastic developed by the Pacific Northwest Laboratory (PNL). This polyurethane-based plastic composition was used as a tissue substitute by Yoder et al. (1979) for direct measurements of the C_x factors using a tissue-equivalent extrapolation chamber (TEEC). Nelson and Chilton (1982) and Nelson (1984) compared results for this plastic and for International Commission on Radiation Units and Measurements (ICRU) tissue using a Monte Carlo technique and reported close agreement. Note that the experimental measurements by Yoder et al. differ from the Monte Carlo calculations for the same phantom geometry, principally at the very low energies but also at mid x-ray energies (see K59 values). The present measurements agree more closely with the Monte Carlo values at the lowest energies and with the TEEC measurements at the mid energies. The K59 and K49 data points are 14% lower than the Monte Carlo calculations. However, the agreement between the Yoder et al. (1979) measurement and the present data suggests that the bias cannot be entirely attributed to the present technique.

The last two columns in Table 3.2 compare the C_x factors specified in ANSI N13.11 (1983) with the present measurements performed on the ANSI N13.11 phantom. The potential for bias not related to this measurement technique is

greater than for the PNL-plastic phantom data because the ANSI N13.11 values are based on Monte Carlo calculations for the ICRU sphere (Dimbylow and Frances 1979). Agreement is good at the lower energies. The discrepancy increases with increasing energy, rising to +11% for the K59 source. The difference in phantom scatter has been documented for the slab geometry compared to the spherical geometry. Nelson (1984) calculated the difference at 60 keV from between +14% and +17%, depending on the tissue constituents used. Ehrlich and Soares (1984) measured the effect of the phantom geometry difference by a TL dosimeter method using PMMA phantoms representing both the sphere and the slab geometries. Results for the shallow depth interpolated to 60 keV indicated a ratio of 1.05 with an uncertainty of 5%. It is reasonable to attribute the 11% overresponse to the difference in geometries. In fact, the agreement between the present measurements on the PMMA phantom and the Monte Carlo calculations for either the PNL-plastic or ICRU-tissue slab phantom is good.

For the bremsstrahlung beams, the agreement between the present measurements and the C_x factors in ANSI N13.11 is good except for the S60 source in which the discrepancy represents about 2.5 standard deviations. The cause of this discrepancy is unclear. The C_x factor for the average energy of the S60 technique (36 keV) estimated from the K-fluorescence data is within one standard deviation for the present measurements. The value from the ANSI N13.11 standard is nearly equal to the value for the K31 source. This is indicative that the discrepancy may not be primarily due to a bias in the present technique. Even though some discrepancies for the body phantom measurements are apparent, the accompanying uncertainties in the correct values for the Monte Carlo and extrapolation chamber techniques implies that the overall bias in the present TL dosimeter technique is less than 15% of the scatter contribution in the worst case (approximately 0.10 in the value for the K59 source). The bias of the TL dosimeter technique is most likely less than this upper limit value.

3.2 EXTREMITY PHANTOM MEASUREMENTS

The measured on-phantom-to-in-air ratios were used in Equation (2.1) to determine the C_x factors. These measured ratios are listed in Table 3.3 along

TABLE 3.3. Measured On-Phantom-to-In-Air Ratios for Extremity Phantoms

Beam	Arm/Leg Phantom						Finger/Toe Phantom			
	A1 Solid	A2 60-mm A1	A3 60-mm BE ^(a)	A4 48-mm A1	A5 48-mm BE ^(a)	A6 25-mm BE ^(a)	F1 Solid	F2 A1	F3 BE ^(a)	F4 Tip
K16	1.11(02) ^(b)	1.02(05)	1.13(02)	1.08(06)	1.09(07)	1.11(02)	1.05(02)	1.09(02)	1.09(02)	1.12(04)
K23	1.14(05)	-	-	-	-	-	1.14(05)	-	-	-
K31	1.25(03)	-	-	-	-	-	1.12(03)	-	-	-
K40	1.41(02)	-	1.20(03)	-	-	-	1.18(05)	-	-	-
K49	1.41(05)	-	-	-	-	-	1.15(04)	-	-	-
K59	1.42(02)	1.23(03)	1.26(03)	1.30(03)	1.32(04)	1.27(03)	1.11(02)	1.13(02)	1.05(03)	1.03(02)
K75	1.36(05)	-	1.17(05)	-	-	-	-	-	-	-
H150	1.24(02)	1.18(04)	1.21(02)	1.20(05)	1.25(04)	1.31(03)	1.08(02)	1.09(02)	1.11(03)	1.12(03)
H250	1.10(03)	-	1.10(03)	-	-	-	1.03(06)	1.02(10)	-	-
¹³⁷ Cs	1.08(03)	-	1.06(04)	-	-	1.09(04)	0.99(04) ^(c)	0.99(03) ^(c)	1.01(03)	0.98(03) ^(c)
⁶⁰ Co	1.01(02)	-	1.06(03)	-	-	1.07(04)	1.02(03)	1.04(02)	1.01(02)	1.02(02)
⁹⁰ Sr/ ⁹⁰ Y	1.01(03)	-	-	-	-	0.97(04) ^(c)	1.02(03)	1.02(04)	1.00(04)	0.99(03) ^(c)
M30	1.17(04)	-	-	-	-	-	1.06(04)	-	-	-
S60	1.30(07)	-	-	-	-	-	1.08(04)	-	-	-
M150	1.39(05)	-	-	-	-	-	1.11(03)	-	-	-

(a) BE is bone equivalent.

(b) 1.11(02) is read 1.11 ± 0.02 .

(c) Ratios less than 1 are attributed to measurement error.

with the estimated one standard deviation uncertainty for the random measurement error. This uncertainty estimate was derived from the variability of the TL readings. Some of the entries represent only one measurement (of at least five dosimeters each for the numerator and denominator of the ratio), while others represent several replicate measurements. The minimum quoted random uncertainty was set at 0.02. The table presents the measured ratios as a function of phantom design and source designation.

The energy dependence of the ratio was determined for the solid PMMA phantoms. The response for the other phantoms was checked for only selected energies. A significant difference was found between the response ratio for the finger and the arm phantoms, as expected. The level of scatter observed for the finger phantoms was sufficiently small so that differences between phantom designs were typically smaller than the uncertainties of the measured ratios except possibly for the K59 technique. The ratio for the finger-tip phantom is significantly (99% confidence level) lower than for the phantoms mounted perpendicular to the beam axis. Significant differences in the scatter response for the arm phantoms were found for energies between 40 and 100 keV.

The response ratios as a function of energy are plotted in Figure 3.2 for the solid arm and finger phantoms and the arm phantom with bone-equivalent insert. The response ratio for the solid plastic arm phantom is systematically greater than the ratio for the phantom with bone-equivalent insert between 30 and 100 keV. This effect is probably a result of the fundamental interactions involved.

The primary process contributing to the backscatter of photons at these energies is Compton scattering. The energy of the Compton-scattered photons is less than the primary-beam energy (approximately 20% less at 60 keV for a backscattered photon). The probability of the photoelectric process increases with increasing atomic number and decreasing energy, increasing the absorption of the Compton-scattered photons in the higher atomic number material. The ratio of the mass attenuation coefficient for 50 keV to that for 60 keV increases from 1.07 for plastic to 1.28 for bone. The deposition of energy per unit mass (dose) is greater for the higher atomic number material because it is more efficient in converting photon energy to electron

FIGURE 3.2. Energy Dependence of the Response Ratios for the On-Phantom-to-In-Air Ratios

kinetic energy, for both the primary and the secondary photon fields. The electrons backscattering from the higher atomic number material are absorbed in the plastic cylinder, so that only the backscattered photon field penetrates to contribute to the surface dose. The greater absorption of the scattered photons in the higher atomic number material decreases the surface dose compared to the dose for the solid PMMA phantom.

The measured ratios for all of the arm phantoms with inserts are similar. The measured ratios for the 60-mm inserts and the 25-mm insert are approximately the same. They all were placed at a depth of 6 mm from the phantom surface. The averaged ratios for the phantoms for the smaller (48-mm) inserts were larger than those for the larger inserts by approximately two standard deviations. The differences observed between the aluminum inserts and the bone-equivalent inserts were not statistically significant. The depth of the higher atomic number insert is more important than its size or composition.

The derived C_x factors are listed in Table 3.4 for the arm or leg phantoms of solid PMMA and of the 60-mm bone equivalent material plus PMMA shell and for the finger phantom of solid PMMA. The C_x values range up to

TABLE 3.4. Exposure-to-Dose Conversion Factors for Extremity Phantoms

Beam	C_x Factor, rad/R		
	Arm/Leg Phantom		Finger Phantom
	Solid PMMA (A1)	Bone Equivalent Insert (A3)	Solid PMMA (F1)
K16	0.97	0.98	0.92
K23	0.99	-	0.99
K31	1.08	-	0.97
K40	1.22	1.05	1.02
K49	1.25	-	1.02
K59	1.28	1.14	0.99
K75	1.26	1.08	-
H150	1.18	1.15	1.02
H250	1.05	1.05	0.99
^{137}Cs	1.04	1.02	0.95
^{60}Co	0.97	1.02	0.98
M30	1.01	-	0.92
S60	1.13	-	0.94
M150	1.27	-	1.02

1.28 for the solid PMMA arm or leg phantom, and up to 1.15 for the arm or leg phantom with bone equivalent insert. The C_x values for the finger phantom are all near unity.

3.3 UNCERTAINTY ANALYSIS

Measurement errors have a random component and a systematic component. Random uncertainties are estimates of the magnitudes of the measurement errors that vary for each measurement performed under the same conditions. Random errors can alter the numerical results if the data were collected again using the same equipment and techniques. Random uncertainties, which are estimates of the probable magnitudes of the random errors, have been estimated from the variability of the TL dosimeter readings and are given in Table 3.3.

The systematic errors are all others that effect the outcome of the measurement. Upper-limit estimates of those factors identified as contributing to the total error are listed in Table 3.5 for three example sources. The total uncertainty of the derived C_x factors can be estimated by calculating the sum of the systematic uncertainty and twice the random uncertainty (95% confidence level) derived from the measured ratios. For example, the total uncertainties at the 95% confidence level for the K16, K59, and ^{137}Cs factors for the solid arm phantom are 4.8%, 4.9%, and 6.9%, respectively. The systematic uncertainty due to the TL dosimeter energy response estimated for the K59 beam (1.8%) represents 6% of the dose attributed to phantom backscatter. This compares to the upper limit of 15% derived from the body phantom measurements. The 6% estimate, based on the effective energy of the back-scattered photons and the energy response of the TL material, is considered a better estimate than the derived limit of 15%.

TABLE 3.5. Systematic Uncertainties

Contributor	Uncertainty Upper Limit Estimate, %		
	K16	K59	^{137}Cs
f-Factor	<1	<1	<1
TL Dosimeter Energy Response	0.2	1.8	0.8
Exposure Delivery	0.4	0.4	0.2
Beam Energy	0.5	0.3	N ^(a)
Total	1.2	2.1	1.3

(a) Negligible.

The systematic errors are all others that affect the outcome of the measurement. Upper limit estimates of these factors identified as contributing to the total error are listed in Table 3.5 for three examples. The total uncertainty of the device C factor can be estimated by calculating the sum of the systematic uncertainty and twice the random uncertainty (95% confidence level) derived from the measured values. For example, the total uncertainty at the 95% confidence level for the K10, K20, and K30 factors for the solid arm position are 1.5, 1.4, and 0.9, respectively. The systematic uncertainty due to the TL dosimeter energy response is listed for the K20 beam (1.8%) representing 5% of the dose attributed to photon dose. Further, this component to the upper limit of 1.52 derived from the body position was subtracted. The 5% estimate, based on the effective energy of the beam was subtracted photon and the energy response of the TL dosimeter, is contained a better estimate than the derived value of 1.52.

TABLE 3.5. Systematic Uncertainty

Contributor	Uncertainty Upper Limit Estimate, %		
	K10	K20	K30
C-factor	1.1	1.1	1.1
TL Dosimeter Energy Response	0.2	1.8	0.3
Exposure Delivery	0.4	0.4	0.1
Beam Energy	0.1	0.3	0.1
Total	1.8	2.6	1.6
(a) Modified			

4.0 CONCLUSIONS

Exposure-to-dose conversion factors were determined for several simple phantom designs by measuring the ratio of the (tissue) dose on the phantom surface to the (tissue) dose in air. The measurement technique chosen using thin LiF TL dosimeters included some systematic error; however, the goal of a total uncertainty of near 5% was achieved for most of the data set. The technique was partially verified through a comparison of measurements on body phantoms to tabulated values from the literature. The value of the comparison was limited because of the uncertainty of the tabulated data and geometry differences between the measurement conditions and the conditions for the tabulated values.

Significant differences in the determined C_x factors were found for the phantoms investigated. The factors for the finger phantoms were uniformly less than for the arm or leg phantoms. The factors for the finger phantoms did not significantly vary with phantom construction because the contribution to the surface dose from the phantom remained small at all energies. The factors for the arm or leg phantoms with either bone-equivalent or aluminum inserts were less than that of the solid PMMA phantom over the energy range from approximately 40 to 100 keV. This effect was attributed to the preferential absorption of the lower energy backscattered photons in the higher atomic number material of the insert. The differences in factors between the phantoms using the bone-equivalent core and the phantoms with the aluminum core were less than the experimental uncertainty at the energies providing the greatest level of backscatter. The observed variation on the size of insert was +8% at 60 keV for the 48-mm diameter compared to the 60-mm diameter inserts.

Due to the small variation of the C_x factor with phantom composition for the finger phantom, the solid PMMA phantom is recommended for use in routine calibrations. Because of the relatively large dependence on phantom construction observed at middle x-ray energies for the arm or leg phantom, a phantom with an insert of higher atomic number material is recommended for use. Closer correspondence between surface dose for the calibration phantom and surface dose on the extremity of the worker for the same radiation field

conditions should result. The position of the material below the surface of the phantom is more important than the size of the insert. For the simplest design, the phantom with the aluminum insert is recommended for use. The diameter of the insert may vary between 48 and 60 mm inclusive, without resulting in a large difference in C_x factors.

The use of the C_x factors given in Table 3.4 is awkward because of the variation of the values attributed to experimental error. The data were used to develop smooth curves as a function of energy, resulting in improved estimates of the C_x factors over the range from 15 to 1500 keV. This data is presented in Table 4.1 for the solid PMMA arm or leg phantom, the 60-mm

TABLE 4.1. Exposure-to-Dose Conversion Factors (smoothed data)

Energy	C_x Factor, rad/R			
	Arm/Leg		Finger/Toe Solid PMMA	ICRU ^(a) Sphere
	Solid PMMA	Bone Equivalent Insert		
15	0.96	0.96	0.92	0.90
20	0.99	0.99	0.95	0.94
30	1.08	1.03	0.98	1.11
40	1.21	1.06	0.99	1.34
50	1.26	1.10	0.99	1.50
60	1.28	1.13	1.00	1.52
80	1.25	1.15	1.01	1.48
100	1.21	1.15	1.02	1.43
120	1.16	1.13	1.02	1.36
150	1.10	1.10	1.01	1.30
200	1.07	1.07	1.00	1.22
300	1.04	1.04	0.99	1.12
400	1.03	1.03	0.99	1.08
600	1.02	1.02	0.98	1.03
800	1.01	1.01	0.98	1.03
1000	1.00	1.00	0.98	1.02
1500	0.99	0.99	0.98	1.00

(a) Dimbylow and Frances (1979).

bone-equivalent center arm or leg phantom, and the solid PMMA finger phantom. The values for the ICRU sphere are also listed for comparison (Dimbylow and Francis 1979). The values over most of the energy range are consistent, with the maximum factors for the solid PMMA arm or leg phantom and the ICRU sphere occurring at the same energy. There is an unexplained difference at the very lowest energies of approximately 5%. Figure 4.1 presents the data with one standard deviation random uncertainties together with the developed smooth curves for the three phantoms.

FIGURE 4.1. The C_x Data with One Standard Deviation Random Uncertainties and Derived Smooth Curves

photo-multiplication factor for the detector and the solid angle of the detector. The values for the LCR are shown in Table 1 and are constant for constant thickness and energy. The values of the energy range are constant with the maximum factor for the solid angle and the LCR. There is an unexpected difference at the very lowest energies of approximately 25 eV. The data points are all within one standard deviation range of the value with the detector. The values for the three points.

FIGURE 1.1 The C Data with One Standard Deviation Random Uncertainties and Derived Factor Curves

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APPENDIX

PROCEDURES FOR USE OF VINTEN THIN TL DOSIMETERS

APPENDIX

PROCEDURES FOR USE OF VINTEN THIN TL DOSIMETERS

Procedures for the use and handling of thin TL dosimeters manufactured by Vinten Instruments were developed to allow annealing and reuse. Element sensitivity correction factors were determined to allow precise measurements to be performed. The procedure follows.

1. Remove the black 7-mg/cm² plastic tape from the dosimeter face and discard the "bandaid" holder.
2. With scissors, trim the Teflon[®] tape from around the dosimeter.
3. Place the trimmed dosimeters in a petri dish face upwards.
4. Place the petri dish (without cover) in a furnace that has been preheated to 275°C.
5. Off-gas and anneal dosimeters for 15 minutes.
6. Remove petri dish and dosimeters from furnace (under reduced light - no fluorescent light) and place in a 100°C preheated oven for 2 hours. This resensitizes the dosimeter phosphor.
7. Place dosimeters in a light-tight, radiation-shielded storage location until thoroughly cooled, preferably for several hours.
8. Fabricate a 5-mg/cm²-thick holder from black polyethylene plastic. The holder is a small pouch large enough to hold one dosimeter.
9. Mount the dosimeters in the holders and label holders to retain dosimeter identification. Expose to 1 rad ¹³⁷Cs at a depth of 1 cm in plastic. Store dosimeters in a dark, shielded storage location until readout.
10. Fabricate a means of holding the dosimeter in a flat restrained position on the TL reader planchet. (A small platinum wire loop was used.)
11. Read out dosimeters under a nitrogen atmosphere, not exceeding 275°C during the readout process.
12. Repeat exposure/readout cycle for the experiment and a post experiment calibration to the ¹³⁷Cs source. (At least six cycles were performed before dosimeter deterioration was observed.)

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