Paper

RECONSTRUCTION OF DOSES FROM OCCUPATIONALLY RELATED MEDICAL X-RAY EXAMINATIONS

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Abstract—**Many nuclear weapons complex workers were required to undergo medical x-ray examinations as a condition of their employment. To ensure that their dose reconstructions are complete, it is necessary to include the contributions from these examinations. X-ray procedures that must be evaluated include: (1) posterior-anterior and lateral radiography, and/or photofluorography, of the chest; (2) anterior-posterior, lateral and oblique lumbar, cervical and thoracic radiography of the spine; and (3) radiography of the pelvis. Each is discussed in the context of conditions that existed during the time the worker was employed. For purposes of dose reconstruction, the x-ray beam size is especially important because the dose conversion factors (DCFs) for each specific body organ depend on whether it was in, or on the periphery of, the primary beam. The approach adopted was to use the DCFs, combined with the entrance kerma, to estimate the organ doses. In cases in which beam output data or information on the primary factors influencing the dose are not available, methods to provide conservative (i.e., claimant-favorable) entrance kerma and dose estimates are adopted. These include specific default values for chest radiography. To account for uncertainties, the estimated doses due to x-ray examinations are increased by 30%.**

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INTRODUCTION

Under the Energy Employees Occupational Illness Compensation Program Act (EEOICPA) (U.S. Congress 2000), x-ray examinations administered in conjunction with routine or special physical examinations required as a condition of employment are considered by the National Institute for Occupational Safety and Health (NIOSH) as a source of occupational exposure. In contrast to the other radiation sources, however, the doses from these sources were neither measured nor included as part of the overall occupational radiation exposure record of the worker, which was in accordance with long-standing practice and recommendations of advisory bodies such as the National Council on Radiation Protection and Measurements (NCRP) and the International Commission on Radiological Protection (ICRP). To complete the exposure record for claims covered under EEOICPA, it was necessary to reconstruct these doses.

Adding to the challenges of reconstructing doses from occupationally required medical x rays was the fact that continuous changes and upgrades have been made in the past 60 years in diagnostic x-ray equipment and the methodologies under which they were applied. Further complicating the situation was that information on the specific apparatus being employed at various nuclear weapons complex sites was meager, particularly during the early years of the U.S. nuclear weapons program. For this reason, accurate reconstruction of these doses has proved to be difficult. Fortunately, changes in the practices and standards recommended during this time period by various organizations, and the conditions under which they were applied, have been documented in a number of publications. Prominent examples are: Brodsky and Kathren (1989); Brodsky et al. (1995); Bushong (1973); Cardarelli (2000); Daniels and Schubauer-Berigan (2005); Ingraham et al. (1953a and b); Kathren and Brodsky (1996); Moeller et al. (1953); Stannard and Kathren (1995); and Taylor (1971, 1979, 1989). Described in the sections that follow is the development of a scientifically based methodology for reconstructing the doses from occupationally required screening x rays; it should be noted that x-ray procedures associated with work-related injury are not included in the dose reconstructions for EEOICPA claims.

TECHNICAL FACTORS AFFECTING DOSE

A number of factors determine the dose due to a medical screening x-ray procedure. For a more or less standard medical radiographic x-ray machine with a

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tungsten target (anode) and focal spot size of 1–2 mm, these include the basic machine settings used for the exposure, namely, the voltage applied to the tube, the current applied to the tube current, the time of exposure, the source (of x rays)-to-skin distance (SSD), x-ray waveform, amount and kind of filtration used, collimation or beam limitation, x-ray tube housing characteristics, the type and speed of the film, development procedure, screens, grids, and the size of the worker being exposed. While the list of factors enumerated may appear to be formidable, in the absence of direct measurements of the beam itself, which are rarely available, the dose to various body organs can be estimated based on knowledge of three basic machine parameters: the applied voltage, tube current, and time (duration) of exposure. The implications of these parameters, insofar as absorbed dose is concerned, are discussed below.

Applied voltage and filtration

X-ray beam energy is characterized by the term *beam quality* and is determined by the applied voltage (peak kilovoltage, kVp) and the amount of added filtration. A medical x-ray apparatus produces a bremsstrahlung spectrum of x rays ranging in energy from near zero to the applied voltage, overlain with the approximately 58 keV K characteristic x rays from the tungsten target. For a typical unfiltered x-ray spectrum, the average energy is about one third of the peak or maximum x-ray energy. Therefore, most of the x rays produced are considerably lower in energy than the applied voltage of the tube, and thus are attenuated by the body and never reach the film. These low-energy x rays are of little value in radiography but contribute significantly to the absorbed dose.

All x-ray tubes have so-called inherent filtration (i.e., the window, aperture, or port in the tube enclosure through which the x-ray beam emerges). In medical radiographic machines, this opening is purposely made very thin to minimize beam attenuation. It is typically equivalent to 0.5 mm of aluminum (Al), in terms of attenuation, and produces little effect on beam quality. To reduce the absorbed dose from this "soft" portion of the energy spectrum, the x-ray beam is hardened by the addition of external filtration (Fig. 1; Ingraham et al. 1953b). Under these conditions, the average energy of the x-ray beam is increased. A corollary to this technique is to apply a higher voltage to the x-ray tube.

Beam energy is typically specified in terms of the half value layer (HVL) in mm of Al. Unfortunately for dose reconstruction, this parameter is seldom specified. Furthermore, even if known, it is of limited value, in part because it does not specify the maximum energy of the beam or its true quality since, in the course of measuring

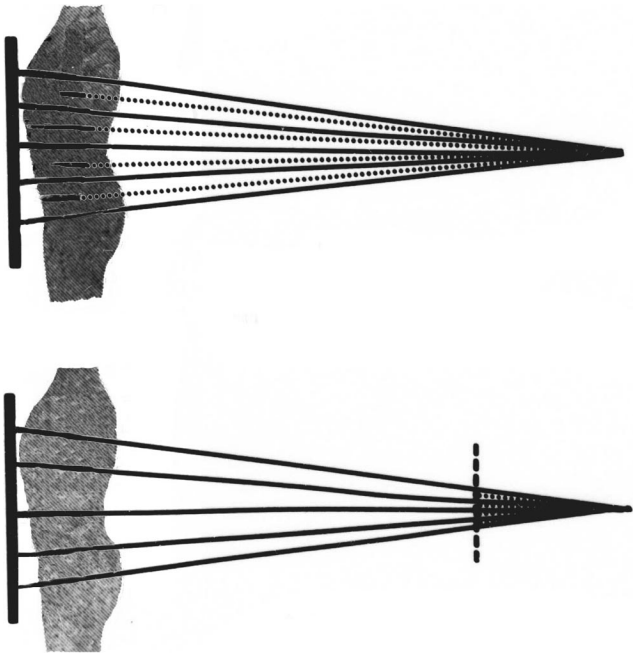

Fig. 1. Effect of external filtration on hardening of the x-ray beam during a radiographic examination of the chest.

the HVL, the absorbers act as filters and the beam is further hardened. Thus, the first HVL is always thinner than the second, which in turn is even less than the third, and so forth. A useful, although rarely available, measure is the homogeneity factor, which is the ratio of the second to the first HVL (Trout et al. 1952). Since the first HVL is always the thinnest, the homogeneity factor will always be greater than 1. The closer this factor approaches unity, the more the beam behaves like a monoenergetic beam.

Although the benefits of filtration with respect to improved radiographic images were known and understood as early as 1896–in fact, within months of the discovery of x rays (Magie 1969), radiographs were made with no added filtration, a practice which quickly became standard and lasted for decades. Recommendations for beam filtration, albeit not specific as to thickness, were put forth in the last half of the 1930's by the International Commission on Radiation Units and Measurements (ICRU 1937). In so doing, the ICRU specified Al filters for x rays of 20–120 kVp, a range that encompassed the voltages in use in diagnostic x-ray units at that time. This was consistent with the 1936 recommendation of the U.S. Advisory Committee on X-Ray and Radium Protection (Table 1), the forerunner of the NCRP, which called for total filtration of 0.5 mm of Al, or equivalent, for medical radiographic machines, and 1 mm of Al for fluoroscopy (NBS 1936). In general, manufacturers of radiographic x-ray machines complied

Table 1. Recommendations of various U.S. organizations on the use of filters in medical x-ray units.

Source of recommendation	Year	Type of unit and equivalent filter thickness
Committee on X-Ray and Radium Protection	1936	Medical radiographic units: 0.5 mm A1 Fluoroscopic units: 1 mm A1[a]
National Committee on Radiation Protection	1949	Radiographic units: 1 mm A1[b]
National Committee on Radiation Protection	1955	New diagnostic x-ray units: 2 mm A1[a]
National Council on Radiation Protection and Measurements	1968	Radiographic units operating above 70 kVp: 2.5 mm A1

[a] Includes both inherent filtration in the x-ray tube housing and that added external to the x-ray tube.
[b] When examining thick parts of the body, such as the chest.

with this standard, and medical radiographic tubes in use in the 1940's typically had inherent filtration of 0.5 mm Al (Morgan and Corrigan 1955).

Typical external or added filtration in the 1940's ranged from none to 1 mm Al. At the latter part of that decade, the NCRP recommended the addition of 1 mm Al filtration for radiographing thick parts of the body, such as the chest (NBS 1949). This thickness had been in prior use in 100 milliampere (mA) units in the larger military hospitals during World War II, and was presumably in use at the various Manhattan Engineer District sites since they were under the aegis of the U.S. Army (Olson et al. 1966). Recommended thicknesses were subsequently increased not only for patient protection but also for improved radiographic image quality. In the mid-1950's, the NCRP recommended that the total filtration—permanently in the useful beam—be equal to 2.5 mm Al (NBS 1955). This was later reconfirmed by the NCRP (1968), the successor to the National Committee on Radiation Protection.

The relationship of *beam intensity* to applied kVp and to filtration for a specific type of x-ray apparatus and x-ray tube is either determined empirically or theoretical values can be obtained from the literature. Since added filtration reduces the entrance skin exposure (ESE)[§] in an exponential manner, these data can be readily modified to account for differences in filtration. For a typical single phase half, full, or self-rectified machine operating in the diagnostic range of 80–100 kVp, each additional mm of Al filtration will effect a reduction of about 40% in the ESE (Trout et al. 1952; Taylor 1957). The approximate intensity reduction afforded by any thickness of Al filtration can thus be determined by the following exponential equation:

$$I = I_0\, e^{-0.4x}, \tag{1}$$

in which x is the thickness of Al, in mm, and I and I_0 are the beam intensities with and without the Al filter present, respectively. In the absence of specific measurements or empirical data, this correction can be applied to determine the effect of filtration on beam intensity.

In contrast to the impacts of filtration, an increase in the voltage applied to an x-ray tube increases the beam intensity. While the increase can be calculated using what is known as Kramer's rule, such calculations are complex and time consuming even with high-speed computers. Fortunately, an alternate method exists to accomplish this task. This is based on the fact that, for a given amount of filtration, the beam intensity increases by the 1.7 power of the applied voltage (Handloser and Love 1951; Trout et al. 1952; Kathren 1965; Cameron 1970). The effects of filtration and voltage tend to offset one another [i.e., added filtration reduces the *exposure* per milliampere second (mAs), while raising the average kVp increases the beam intensity]. Higher kVp radiographic techniques typically require shorter exposures in terms of mAs, and the dose reduction from additional filtration, at the recommended level, more than offsets the additional exposure from using increased kVp. However, there is not a direct correlation or proportionality between the effects of filtration and kVp; thus, corrections for each parameter should be independently determined or can be computed using the following equation:

$$I = I_0\, e^{-0.4x} \left(\frac{\mathrm{kVp}}{\mathrm{kVp}_0}\right)^{1.7}, \tag{2}$$

where I is the beam intensity at a newly applied kVp, I_0 is the beam intensity at the reference or kVp_0 originally applied, and x is the thickness of the added filter in mm Al.

Collimation

Radiographic x-ray tubes are typically enclosed in lead shielded protective housings designed to limit the leakage of radiation from the x-ray tube to the recommended rate of less than 0.1 R (~1 mGy) h^{-1} at 1 m distance, with the primary or useful beam exiting through

[§] ESE actually refers to the *exposure* at the point where the skin of the patient is closest to the target and does not account for x rays that are scattered from the patient. With later definitions of *exposure* and radiological quantities, air kerma has been used in place of ESE. However, both quantities are essentially the same, at least conceptually, for purposes of dose reconstruction.

a port or window in the side of the housing. This obviates any need to consider radiation leakage from the tube as a contributor to exposures in the dose reconstruction process. The size of the beam port, and its distance from the focal spot (i.e., the point of production of the x rays), determine the size of the beam at any distance from the tube. In a manner analogous to the previous discussion of x-ray filtration, these factors may be considered sources of the inherent collimation of the x-ray beam. It is common to add features to further restrict the beam size so as to reduce both the scattered radiation and accompanying absorbed dose to the individual. Otherwise, organs normally outside of the primary beam, and of no interest radiographically, are unnecessarily exposed. Even though it is known that features to collimate the beam were widely used during the years prior to 1970, data necessary to estimate the size of the beam may not be available. As a result, it is standard practice for the purposes of dose reconstruction to assume that minimal or no additional external collimation was used in the absence of information to the contrary. In cases where the size of the beam port and its distance from the focal spot are not known, the default assumption is that the beam port is 6 cm in diameter and located 5 cm in front of the focal spot. These dimensions are consistent with the physical characteristics of x-ray tubes in use at that time, and they are sufficiently conservative to ensure that the resulting dose estimates will be favorable to the claimant.

Other factors

A number of other factors may also increase the exposure of the worker to x rays and the accompanying absorbed dose he or she receives. As noted above, knowledge of these factors is not necessary for dose reconstruction purposes if beam measurements are available, or if the primary machine characteristics of applied kVp and current and time (mAs) are known along with the amount of primary beam filtration. These include the size of the individual being x-rayed, which may need to be taken into account on a case-by-case basis; appropriate factors to account for patient size are provided in Table 2. For some procedures, such as chest photofluorography (PFG) in which the exposure is regulated by the exposure to the film, the apparatus automatically compensates for patient size or thickness, and no adjustment or corrections are needed. The exposure needed for a suitable radiograph is a function of film speed and development methods that have changed over the years. However, the effects of film speed and development do not need to be considered, since dose reconstruction is based on the ESE.

There is evidence that the Potter-Bucky grid system was used for posterior-anterior (PA) chest examinations at various U.S. Department of Energy (DOE) and Atomic Weapons Employer (AWE) facilities and sites, the reason being that the additional attenuation of the beam improves the quality of the radiograph and makes it a better diagnostic tool (Sante 1954; Thomas et al. 1959). If actual measurement data are not available for purposes of dose reconstruction, the dose can be reasonably estimated by multiplying the doses without the Potter-Bucky grid system by a factor of 2.5. A summary of the effects of the technical factors discussed above is shown in Table 2.

HUMAN FACTORS AFFECTING DOSE

In addition to the technical factors discussed above, human factors played a role in the doses received by workers due to diagnostic x-ray examinations—retakes were primarily due to erroneous perceptions. Trout et al. (1973), based on an analysis of the rejection rate of chest radiographs obtained during the Coal Mine "Black Lung" program (U.S. Congress 1969), reported an average rejection rate of 3% among 67,000 radiographs. The retake rate in the DOE system was likely much smaller, one reason being that a high percentage of the examinations were of the chest, a well-established and standardized procedure. A few retakes may have occurred in females to account for breast tissue; in African-Americans to account for the historically perceived higher density of their bones; and in the administration of x-ray examinations in larger individuals using technique

Table 2. Relationship of various technical factors to the intensity of the x-ray beam.

Parameter	Units	Relationship to x-ray beam intensity
Applied voltage	kVp	Intensity proportional to 1.7 power of kVp
Tube current	mA	Linear
Exposure time	s	Linear
Filtration	mm A1	Intensity decreases by ~40% for each added mm of A1
Patient size (chest thickness—applicable to chest films only)	25–27 cm	Dose increased by factor of 1.5
	>27 cm	Dose increased by factor of 2
Distance	d	$1/d^2$ at $d \geq 30$ cm from the x-ray tube target
Potter-Bucky grid system		Entrance skin exposure (kerma) is assumed to be increased by factor of 2.5 if usage and other data are not available

factors for smaller individuals. Without detailed information on the machine settings that were used, it is not possible to estimate the increase in dose that occurred. As will be noted in the section that follows, the dose from fluoroscopic procedures was also significantly influenced by the application of poor or improper techniques.

TYPES AND FREQUENCIES OF MEDICAL X-RAY EXAMINATIONS

Since there was no standard practice with respect to the x-ray screening procedures that were required by the contractors at various DOE nuclear weapons sites, either in terms of pre-employment practices or for workers already on the job, the types and frequencies of required medical x rays were variable and generally site specific. Another complicating factor was that the nature of the examinations changed with time. For example, required medical x rays were more common and frequent in the early years (i.e., the 1940's and 1950's) and less in later years, in accord with the typical pattern of medical x-ray screening practices reported in the literature and cited in the Introduction.

Chest radiography and PFG

As a general rule, pre-employment chest x-ray examinations were required at most AWE facilities and DOE nuclear weapons sites, along with an annual or biennial follow-up performed in conjunction with a routine physical examination. Practices at the Hanford Site (Richland, WA), for which relatively complete documentation is available, indicate that during the early years workers identified as "at risk" were given medical examinations, including x rays, at more frequent intervals than other workers (Cantril 1951). For work involving exposures to radiation, the interval between examinations in January 1944 was as close to 4 wk as possible. By July 1945, the intervals between examinations were increased to 7 or 8 wk. In contrast, workers not involved with radiation or other special hazards were examined every 3–6 mo (Cantril 1946). It was not until 1959, when the schedule for such examinations was based on age with an accompanying limit of no more than one per year, that a significant reduction in their frequency occurred.

In the case of chest x-ray examinations, the procedure was usually limited to a single PA chest radiograph, although a lateral chest radiograph might also have been taken. Stereo chest films also may have been required for some workers; this procedure required two separate exposures with the views slightly displaced. At some sites, workers were examined with chest PFG units. This procedure (also known as photoroentgenography) was unique in that the image was displayed on a fluoroscopic screen and photographed; this then served as the x-ray image (Fig. 2; Ingraham et al. 1953b). While PFG units were conservative in terms of the use of film, the accompanying dose was much higher than that required by a standard PA radiograph. In fact, the absorbed dose to the chest was estimated to be about 1 R (~10 mGy) (Moeller et al. 1953). For these and other reasons, these units had been phased out by 1970 and, in most cases, a few years earlier.

Fluoroscopy

Fluoroscopy, in contrast to PFG, involves real time viewing of a fluorescent screen continuously activated by x rays. Because of the time required and other limitations, this technique was not generally amenable to mass examinations or to pre-employment screening of workers and was not a recommended practice as a substitute for radiography (Rigler 1938, 1954; Hodges et al. 1947; De Lorimer et al. 1953; Rabin 1968). Despite the admonitions to the contrary, there are indications that fluoroscopic examinations of the chest were conducted and required at least at one site (Linde Ceramics, Tonawanda, NY) during 1942 and 1943, and it is possible that such examinations also were performed at other sites. The ESE of such units were high but variable, ranging from 2 R (~20 mGy) min^{-1} for a well-operated "modern" unit with appropriate filtration, to several tens of R (~tenths of Gy) min^{-1} for older, poorly maintained units with short target-to-panel distances and perhaps inadequate filtration.

Another human factor that significantly influenced the dose was the exposure time, which was dependent not only on the machine output but also on the techniques of

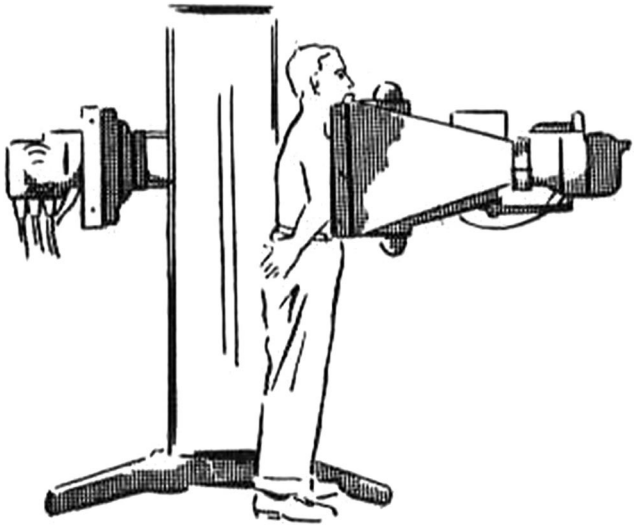

Fig. 2. Schematic drawing of the components of a photofluorographic x-ray machine.

the radiologist. Some radiologists applied the x rays in short bursts, others just "put their pedal to the metal;" that is, they basically kept their foot on the x-ray tube operating switch throughout the entire procedure. Also contributing to increased doses was the failure of the radiologists to provide adequate time for adapting their eyes to the dark, which necessitated longer exposure times and increased patient dose. Therefore, many machines were equipped with timers to restrict the exposure, normally to 25 R (~0.25 Gy) at the panel, for any given examination. In most cases, the total exposure time was probably less than 1 min—perhaps 15–30 s—although exposures of a minute or more were not uncommon. To estimate doses as favorable to the claimant, a high but not unreasonable exposure time of 2 min and an ESE rate of 20 R (~0.2 Gy) min^{-1} are assumed in the absence of data to the contrary.

Spinal and pelvic radiography

At some sites, lumbar spine radiographs were routinely required for certain classes of male workers to screen for the presence of back problems. The frequency and number of lumbar spine views were site dependent. Examinations for evaluating back problems likely included both an anterior-posterior (AP) and lateral view, and possibly an AP and lateral spot film. In the absence of data, it was assumed that all views—a total of four (two AP and two lateral)—were taken. Recommended practice was to use a 5-inch-diameter (12.7-cm) cone (Sante 1954) for improved radiographic quality, thus limiting the beam diameter to 12.7 cm at the point of skin entrance.

X-ray examinations of the pelvis, or cervical and thoracic spine, were also used in the 1940's for medical monitoring of workers with the potential for occupational exposure to fluoride and fluoride compounds (Osinski 1947; Key et al. 1977). By 1970, more specific and sensitive screening methods were available. Therefore, it is reasonable to assume that radiographic diagnosis of fluorosis was discontinued after that date.

ORGAN DOSE RECONSTRUCTION METHODOLOGY

Organ dose reconstruction is conceptually a relatively simple two-step process for x-ray examinations. The first is to determine the entrance kerma; the second is to determine the specific organ doses. The latter step is generally performed using the standard conversion data provided in ICRP Publication 34 (1982). This methodology is based on elaborate Monte Carlo calculations for detailed anthropomorphic models, largely derived from the work of Kereiakes and Rosenstein (1980). Following this approach, organ dose (OD) is obtained as the product of entrance kerma (EK) and the associated dose conversion factor (DCF):

$$OD = DCF \times EK. \quad (3)$$

DCFs depend on the x-ray projection, the organ, and the beam quality (expressed as the HVL), and are given in terms of average absorbed organ dose (mGy) per Gy of entrance kerma, defined by the ICRP (1982) as "air kerma in air without backscatter."

Determining entrance kerma

Entrance kerma is best determined from actual measurements of beam intensity or ESE. Use of actual measurement data is the simplest and most direct starting point for assessing x-ray doses; it typically requires few, if any, assumptions, and has the least amount of uncertainty and, hence, is preferred for x-ray dose reconstruction. Where measurements are unavailable, x-ray dose reconstruction can be accomplished using eqns (2) and (3). Further, if both measurement data and technique factors are unavailable, then organ dose estimates can be made using ESE values derived from various sources (Table 3). The numbers presented in the Table are conservative and should be used only as a last resort.

Entrance kerma when beam output measurements are available

Beam output measurements have historically been made in terms of *exposure* and quantified in units of the roentgen (R). Depending on the measurement device and technique, these may have a wide range of uncertainty, with the best measurements made with integrating ionization chambers designed specifically for medical x-ray applications. Until about 1970, there were two such instruments in common usage in the United States, the Victoreen R-meter, and the Landsverk L series ion-chambers. For measurements made with R-meters and similar ion chambers, the associated uncertainty in the energy region of interest generally should not exceed ±2% (Kathren and Larson 1969).

Other integrating devices, including photographic emulsions, pocket ionization chambers, and thermoluminescent dosimeters, have also been used for beam output

Table 3. Default entrance kerma values for common x-ray screening procedures.

Period	Entrance kerma (mGy)		
	PA chest	Lateral chest	Photofluorographic chest
Pre–1970	2.0	5.0	30
1970–1985	1.0	2.5	n/a[a]
Post–1985	0.5	1.3	n/a[a]

[a] Photofluorographic units were no longer in use after 1970.

measurements. Data based on measurements made with these types of dosimeters should be used cautiously, as all are energy dependent, and correction for beam energy is a likely necessity. Beam output measurements usually define or directly determine the ESE, or can be corrected to obtain a reasonable estimate of the ESE for a given procedure by using the generic intensity relationships shown in Table 2. The ESE will, at least for older measurements, be in units of R, which must be converted to kerma and then to organ dose, as shown in eqn (3). The relationship between *exposure* and kerma is given by

$$D = f \times R, \tag{4}$$

in which D is the kerma, R is the *exposure*, and f is constant and somewhat energy dependent. If D and R are expressed in terms of the old units, namely rad and R, respectively, $f = 0.93$. For dose reconstruction purposes, and to provide conservatism, an exposure of 1 R may be taken to be exactly equal to a kerma of 1 rad (10 mGy).

Entrance kerma from technique factors

To obtain estimates of ESE or entrance kerma, the basic data required are kVp, filtration, exposure (mAs), and distance. Beam output data are also available from a number of publications, including McCullough and Cameron (1970) and NCRP Report No. 102 (1997). Table B.3 of Report No. 102** is reproduced below in its entirety as Table 4. As noted, it provides average air kerma rates for medical diagnostic x-ray equipment operating at various kVp with 2.5 mm Al filtration at distances from 30–183 cm from the source.

Corrections for different thicknesses of Al filtration can be made by reference to Table 3.1 in NCRP Report No. 102 (1997). As an alternative, Fig. B.1 in this same reference provides a graphical representation of air kerma at 100 cm for various values of kVp and filter thickness greater than 2.5 mm Al (NCRP 1997). A similar set of curves along with a set of curves plotting the output as a function of focus (target)-to-skin distance for various kVp with 2.5 mm filtration is also provided in pages 159–160 of the 1970 *Radiological Health Handbook* (Cameron 1970); also highly useful is the nomograph from the earlier (1954) version of the *Handbook* (Kinsman 1954). Using these tables or graphs, a reasonable estimate of beam output and hence entrance kerma can be obtained.

** NCRP Report No. 102 was published in 1989. Table B.3 was revised in a 1997 corrigendum.

Default entrance kerma values for chest radiography

As previously noted (Table 3), default values of entrance kerma have been provided for reconstruction of dose for the three most commonly used medical x-ray screening procedures: PA chest radiography; lateral chest radiography; and PFG chest films, when applicable. In determining these default values, it was assumed that a minimum of filtration was used along with low voltage techniques, slow film speeds with standard development, and no additional collimation or use of cones. Similar values for other screening procedures can be developed using technique factors given in the literature of the time.

Converting ESE to organ dose

Once the ESE has been converted to entrance kerma, doses to a number of different organs from various radiographic procedures can be obtained from Tables A2–A8 in ICRP Publication 34 (1982). The entrance skin dose is readily calculated as the product of the entrance kerma times a backscatter factor that can be obtained from Table B.8 in NCRP Report No. 102 (1997), and is applicable to all skin surfaces in the primary beam on the entrance side of the body. Since these two reports serve as the basis for most of the discussion that follows, they will subsequently be cited simply in terms of the ICRP and NCRP tables or reports.

Use of the ICRP tables requires knowledge of the x-ray beam quality and beam intensity. If the kVp and filtration are known, the HVLs can be estimated from the data provided in either ICRP Table A16 or NCRP Table B.2. In general, the higher the kVp and filtration is, the higher the HVL. If the actual beam quality is unknown, a higher rather than a lower HVL should be assumed. This, in turn, likely will result in an overestimation of the organ dose. In the absence of data, the recommended values for beam quality are 2.5 mm Al for radiographs taken prior to 1980, and 4.0 mm Al for subsequent radiographs.

The previously cited Tables A2–A8 do not include all the organs that have been identified in the Interactive RadioEpidemiological Program code, which is used to calculate the probability of causation of radiogenic cancers (Kocher et al. 2008). For those organs that are included but not specifically identified by the ICRP, it is suggested that the DCF for the organ that is anatomically the closest to the one of interest be used. In the case of chest radiography, this would mean that the DCF for the lung would be applied to all other organs within the thoracic or abdominal cavity that may be intercepted by the primary x-ray beam (i.e., the thymus, esophagus,

Table 4. Average air kerma rates produced by diagnostic x-ray equipment.[a]

Distance from source to point of measurement (cm)	Tube potential (kVp)											
	40	50	60	70	80	90	100	110	120	130	140	150
	mGy per 100 (mAs)[b]											
30	19	35	53	72[c] (42)	92 (54)	114 (67)	137 (81)	161 (95)	187 (109)	213 (126)	240 (141)	269 (158)
45	8.4 (4.9)	16 (9.2)	23 (14)	32 (19)	41 (24)	51 (30)	61 (36)	72 (42)	83 (49)	95 (56)	107 (63)	120 (70)
60	4.7 (2.8)	8.7 (5.2)	13 (7.8)	18 (1.1)	23 (14)	28 (17)	34 (20)	40 (24)	47 (28)	53 (31)	60 (35)	67 (40)
100	1.7 (1.0)	3.1 (2.0)	4.7 (3.0)	6.5 (4.0)	8.3 (5.0)	10 (6.0)	12 (7.0)	14 (9.0)	17 (10)	19 (11)	22 (13)	24 (14)
137	1.1 (0.5)	1.7 (0.9)	2.5 (1.5)	3.4 (2.0)	4.4 (2.6)	5.5 (3.2)	6.5 (3.8)	7.7 (4.5)	8.9 (5.2)	10 (6.0)	12 (6.8)	13 (7.6)
183	0.5 (0.3)	0.9 (0.5)	1.4 (0.9)	1.9 (1.1)	2.4 (1.5)	3.1 (1.8)	3.7 (2.2)	4.4 (2.5)	5.0 (3.0)	5.8 (3.4)	6.5 (3.8)	7.2 (4.3)

[a] This table was reconstructed from Table B.3 (corrigendum), NCRP Report 102 (1997). This table, in turn, was based on a Fig. 3 in Zamenhof et al. (1987).
[b] Air kerma values are for total filtration equivalent to 2.5 mm Al.
[c] Values not in parenthesis are for three-phase generators; those in parenthesis are for single-phase generators.

stomach, and liver/gall bladder/spleen). Since an appreciable fraction of the skeleton, in particular, the trabecular bone (which has a large surface-to-volume ratio), and the sternum (which is a primary location of the red marrow in the adult), lies within the trunk, the DCF for the lung would also be applied to the bone surfaces. On this same basis, the DCF for the ovaries would be used for organs in the lower abdomen (i.e., the urinary bladder, and colon/rectum). For the eye and brain, the analogous organ is the thyroid. Furthermore, the organ dose for skin can be obtained by reference to NCRP Table B.8, which provides backscatter factors for different beam qualities and field sizes.

The DCFs for those organs that are nearby and also outside the primary beam can be used as analogues, given their anatomic location and shorter distance from the edge of the modified primary beam. The dose to these organs is assumed to be 10% of that for organs in the neighboring region inside the beam. For other organs (e.g., the ovaries and testes, which are outside and well away from the modified primary beam), DCF values, based on for thoracic and cervical spine procedures, can also be obtained from the ICRP tables. However, it should be noted that calculated doses for these organs, and specifically those for the gonads, are much smaller than the measured values reported in the literature of the time (Braestrup and Wyckoff 1958; Lincoln and Gupton 1957; Norwood et al. 1959; Stanford and Vance 1957; Webster and Merrill 1957). This is probably because more scattered radiation is produced with a larger x-ray field, such as that of a modified primary beam, than with the properly collimated beam assumed by the ICRP. Therefore, to ensure that doses for the gonads are not underestimated, higher measured values are used.

The methodology used to determine DCFs for cervical and thoracic radiography was also used to calculate the DCFs for pelvic radiography. For example, the primary beam was assumed to have been poorly collimated and circular, with an area twice the area of the film and possibly displaced upward or downward by ± 3 cm from the location assumed by the ICRP. To identify additional organs that would be impacted by the enlarged beam, a transparency with an outline of the beam cross-section, properly scaled in size, was overlaid on anatomical drawings of that portion of the body. Scattered radiation, as well as the estimated fraction of the organ volume in the modified direct beam, was considered in developing the DCFs for organs lying outside the area of the collimated beam, but within, or partially within, the assumed modified (i.e., poorly collimated) beam.

A larger beam would also produce a higher dose to the active bone marrow. This was accounted for by using data in Table 116 ("Active Bone Marrow Dose as a Function of Field Size") reported by Kereiakes and Rosenstein (1980). The ICRP active bone marrow DCF for an AP pelvis radiograph was multiplied by 1.44 based on the upper limit of the correction factor 1.39 ± 0.05 given in Table 116 for a 14-inch $\times$ 17-inch AP abdominal radiograph and a beam-to-film area ratio of 2.0.

UNCERTAINTY ANALYSIS

Error implies knowledge of what the correct or actual value is, which is, of course, not known. Therefore, the more appropriate factor is uncertainty, which is expressed in terms of a confidence level (CL), and expressed as a percent. Thus, the 99% CL indicates that the correct or true value, although not actually known, has a 99% probability of falling within the range cited. The CL typically includes all potential sources of error, both random and systematic; the precision or reproducibility of the measurement; and accuracy, or how close the measurement or estimate of dose comes to the actual or correct value.

In theory, a large number of parameters can introduce uncertainties or affect the intensity of the x-ray machine output beam and, consequently, the dose to the

person being examined. In practice, however, the following five basic parameters can be considered to have a meaningful or significant impact on dose as error or uncertainty related to diagnostic medical x-ray examinations:

- Measurement error;
- Variation in applied voltage (kVp);
- Variation in beam current (mA);
- Variation in exposure time (s); and
- Source-to-skin distance (SSD).

The uncertainty associated with each of these parameters is discussed below. The estimated uncertainty assumes that large systematic errors, such as a consistent 25% high reading on a meter or a calibration calculation error that results in a large systematic error, are absent. The influence of other systematic parameters, such as the use of screen, grids, reciprocity failure, and film speed and development, while potentially variable, do not affect the beam quality and intensity, per se, except indirectly, insofar as these may determine the x-ray tube exposure settings (i.e., kVp, mA, and time):

- Medical x-ray doses were largely derived from measurements. As previously discussed, if properly calibrated and applied, R-meters and similar instruments typically and historically have had an uncertainty of ±2% for photon energies below 400 keV (Kathren and Larson 1969). Although more recent versions of these instruments might provide an uncertainty of perhaps half this value (NBS 1985, 1988), for conservatism, the uncertainty range of ±2% will be assumed;
- Ideally, for a given set of machine settings and parameters, x-ray output should be constant and unvarying. This, however, is not true in practice. Output is essentially constant unless focal spot loading occurs, as might be the case when the power rating of the machine is exceeded. This, however, is unlikely because such an event would severely damage the tube. Data show that, for a given kVp setting, the variations generally remain within ±5% (Seibert et al. 1991). Since, as noted earlier, the beam intensity is proportional to the 1.7 power of the applied voltage, this translates to an uncertainty of ±8.6% with respect to output beam intensity in the 80–100 kVp range used for chest radiographs. This is rounded up to ±9%;
- Variations in tube *current* are inevitable. As a tube ages, or heats up from use, the amperage can change, generally in a downward direction. With all other factors constant, beam intensity will be reduced in direct proportion to the change in tube current. Under normal operating conditions, however, the reduction in beam output from current variation is generally within the range of ±5%;
- In contrast to the impacts of the above parameters, the time of exposure is far more significant in terms of medical x-ray dose reconstruction. This is underscored by noting that virtually all diagnostic medical x-ray units used in the DOE nuclear weapons complex were full-wave-rectified, meaning that they produced 120 pulses of x rays s^{-1}. Thus, in a typical radiographic exposure time of 0.05 s, only six pulses would occur. A small error (variation) in the timer that resulted in a change of only ±1 pulse would correspondingly affect the output by ±17% for an exposure time of 1/30 s. The corresponding change in output would be ±25%. Early mechanical timers were notoriously inaccurate; accuracy improved significantly with the introduction of electronic timers. Although measurements of the reproducibility made in the late 1980's and beyond by the Washington State Department of Health (WDH) for the x-ray units at the Hanford Site, for example, suggest that the timers, and indeed the entire x-ray output, were fairly constant (WDH 1990–1999), for conservatism it is assumed that the accompanying uncertainty in beam output attributable to timers was ±25%; and
- SSD, the last parameter, can also affect worker dose. For a given individual, the SSD will be determined largely by his or her body thickness and the accuracy of the exposure position. Normally, the estimated variation in SSD is no more than a few centimeters, with an upper limit of perhaps 7.5 cm. On the basis of the inverse square law, this indicates an uncertainty of ±10% from this source.

The total uncertainty, expressed as propagated error, would be the root mean square value of the individual errors for each of the five parameters. The combined error (σ) can be formulated as follows:

$$\sigma(\%) = \sqrt{2^2 + 9^2 + 5^2 + 25^2 + 10^2} = \pm 29. \quad (5)$$

Therefore, the estimated doses for all x-ray examinations used in the radiation dose reconstruction program were increased by 30%.

COMMENTARY AND CONCLUSION

In contrast to the approach traditionally applied in assessing doses for purposes of occupational exposures, the regulations that apply to the NIOSH Radiation Dose Reconstruction Program require that doses due to work-related medical x-ray screening examinations be included in the dose assessments (Neton et al. 2008). This is particularly challenging because records of such exposures, especially during the earlier years of operation of AWE facilities and DOE weapons complex sites, are

often incomplete, nonretrievable, or were recorded in units that are no longer in use. Although it was a mark of progress, such assessments also are complicated by the continual replacement and upgrading of the x-ray equipment through the years. This included the replacement of older equipment with newer units that had improved collimation, filtration, and timers, and were operated at higher voltages. Another change was the use of faster films and improvements in associated procedures for their development, all of which led to reductions in the doses being received.

Other factors that influenced the doses from x-ray examinations were a reduction in the perceived value of specific x-ray screening procedures and the need to reduce costs. These, in turn, led to a reduction in the frequency and types of x-ray screening examinations that were required. While chest radiographs remained a common procedure applied to most workers, only a very small fraction of them at a limited number of sites were believed to have continued to be subjected to higher dose associated procedures such as, for example, lumbar spine x rays to screen for potential back anomalies. Nonetheless, x-ray examinations of the pelvis, thoracic spine (dorsal), lumbar spine, and cervical spine may have been performed at some sites to screen for occupational skeletal fluorosis in workers with the potential for exposures to fluoride.

To ensure that the medical x-ray portion of the dose reconstruction is complete, procedures and site-specific technical basis documents were developed to include the contributions from screening examinations. This was accomplished by performing a careful evaluation of common x-ray procedures, such as (1) PA and lateral radiography, and/or PFG of the chest; (2) AP, lateral and oblique lumbar, cervical and thoracic radiography of the spine; and (3) radiography of the pelvis, and fluoroscopy of the abdomen and the lower body. Also factored into all assessments were the quality and intensity of the x-ray beam, which played a major role in the accompanying doses. To ensure that the estimated doses were favorable to the claimant in cases in which data were not available, default values which were known to be conservative were adopted. In addition, the estimated doses were increased by 30% to account for the combined uncertainties in the five identified most important instrumentation and operational parameters [i.e., measurement (calibration) error; and variations in the applied voltage, beam current, exposure time, and SSD].

While the scientific aspects of how doses from medical screening examinations are being assessed are of interest, not to be overlooked is the fact that this represents one more step that is taken to ensure that DOE facility workers who developed cancer and submitted claims for compensation are provided the "benefit of the doubt."

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