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Draft	10/27/2003	00-A	New document to establish the technical basis for the development of a generic document to use to perform dose reconstruction from occupationally related diagnostic X-ray procedures. Initiated by Vern Shockley.
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ACRONYMS AND ABBREVIATIONS

cm	centimeter
DF	dose conversion factor
EEOICPA	Energy Employees Occupational Illness Compensation Program Act of 2000
ESE	entrance skin exposure
Gy	gray
HVL	half value layer
ICRP	International Commission on Radiological Protection
ICRU	International Commission on Radiological Units
IREP	Interactive RadioEpidemiological Program
kVp	Peak Kilovoltage, applied kilovoltage
lat	lateral
mA	milliampere
mAs	milliampere-second
mm	millimeter
NCRP	National Council on Radiation Protection
NIOSH	National Institute for Occupational Safety and Health
PA	posterior-anterior
PFG	photofluorography
R	roentgen
RMS	root mean square
SID	source to image distance
SSD	source to skin distance

DOSE RECONSTRUCTION FROM OCCUPATIONALLY RELATED DIAGNOSTIC X-RAY PROCEDURES

Ronald L. Kathren, Vernon E. Shockley, Elyse M. Thomas, and Timothy D. Taulbee

1.0 Introduction

An additional contribution to occupational radiation exposure of workers may be from medical diagnostic x-ray procedures that are imposed upon the worker as a condition of employment. Although clearly occupationally related, the dose from these exposures was typically not measured nor was it considered or included as a part of the overall occupational exposure of the employee. With the passage of the Energy Employees Occupational Illness Compensation Program Act of 2000 (EEOICPA), diagnostic medical x-rays administered in conjunction with routine or special physical examinations required as a condition of employment are recognized as a valid source of occupational exposure and are to be included in the determination or reconstruction of the dose to the worker. Unlike occupational exposures incurred during normal work processes, diagnostic medical x-ray exposures were not monitored, necessitating reconstruction of the doses acquired in this manner.

The EEOICPA is codified in 42 United States Code 7384-7385 and provides compensation for workers and former workers in the nuclear weapons production programs of the U.S. Department of Energy and predecessor agencies (DOE) who have been diagnosed with cancer providing that the cancer was “at least as likely as not” to have been attributable to the occupational dose acquired while working in the DOE programs. To enable this determination to be made, the National Institute for Occupational Safety and Health has been charged under the provisions of Title 42, Code of Federal Regulations, Part 82, with developing dose reconstruction methods and with applying the doses to determined the probability of causation -- i.e. the likelihood that the cancer was attributable to the occupational radiation exposure incurred by the worker. Of necessity, this requires knowledge of specific organ doses. Accordingly, a comprehensive guidance document for external dose reconstruction has been developed to assist qualified health physicists in implementation of the EEOICPA (OCAS 2002). This report supplements and expands upon the guidance provided in the OCAS document by providing more specific and detailed methodology for dose reconstruction from diagnostic medical x-rays that were sustained by workers as a condition of employment, and provides the technical basis for dose reconstruction in the absence of specific dose measurements or records of technique factors. The additional guidance provided in this report is needed because of the paucity of available technical data and records specific to medical diagnostic exposures experienced by workers in the DOE weapons program

2.0 Technical Factors Affecting Diagnostic X-Ray Dosage

A number of factors determine the dose to the patient from a diagnostic x-ray procedure. For a more or less standard medical radiographic (i.e. diagnostic) unit with a tungsten target (anode) and focal spot of 1-2 mm, these include the basic machine settings used for the exposure, viz. the applied kilovoltage of the beam (kVp, also known as peak kilovoltage or kilovolt peak), beam current (mA), and time of exposure, distance, waveform, amount and kind of filtration used, collimation or use of diaphragms, tube housing characteristics, the type and speed of the film, development procedure, screens, grids and the size of the patient. While the list of factors enumerated looks formidable, in the absence of direct measurements of the beam itself, which are rarely available, the dose to the patient can be estimated with a reasonable degree of accuracy with knowledge of only the three basic machine parameters: applied kilovoltage, current, and time, along with filtration, collimation and waveform characteristics. The implications of these factors insofar as patient dose is concerned are briefly discussed below.

2.1 APPLIED KILOVOLTAGE AND FILTRATION

The energy of the x-ray beam is determined by the applied kilovoltage and the filtration, and is sometimes referred to as beam quality. X-rays, as produced in a typical medical x-ray tube, are bremsstrahlung produced when electrons from the cathode are accelerated into the anode as a result of the potential difference or applied kilovoltage between the two electrodes. As such, x-rays from a medical x-ray tube are a distribution or spectrum of energies ranging from zero to the applied kilovoltage, which refers to the potential between the anode and cathode of the tube. For a typical unfiltered x-ray spectrum, the average energy is about one third of the peak or maximum x-ray energy, or applied kilovoltage. Hence most of the x-rays produced are very much lower in energy than the applied kilovoltage of the beam, and thus are attenuated by the torso or other portion of the body being radiographed and never reach the film. These x-rays are of little value in radiography but contribute significantly to patient dose.

To reduce the dose to the patient, filtration in the form of a specified thickness of absorbing material is added to the beam. This has the net effect of absorbing a large fraction of the lower energy x-rays that are of little or no value in making the radiograph while allowing a greater fraction of the more energetic and radiographically useful x-ray photons to pass. In this manner, the dose to the patient is significantly reduced while at the same time radiographic quality may be enhanced. A filtered x-ray spectrum has a correspondingly higher average energy than before it was filtered, although the photon fluence rate is much reduced. Such a beam is said to have been hardened. A corollary to this filtration technique is to use a higher applied kilovoltage, and filter the beam relatively heavily to eliminate most of the low energy radiographically useless photons from reaching the patient.

Beam energy is specified in terms of quality, or hardness, which in turn may be specified in terms of the half value layer (HVL) in aluminum. Unfortunately, this parameter is seldom available, and even if known is of limited value, in part because it does not specify the maximum energy of the beam or its true quality, since as the HVL measurement is made, the absorbers act as filters and the beam is further hardened. Thus the first HVL is always smaller than the second HVL beam, which in turn is smaller than the third, and so forth. A useful although rarely available measure is the homogeneity factor, which is simply the ratio of the first and second HVLs. Since the first HVL is always the smallest, the homogeneity factor will always be < 1, and the closer it approaches unity, the more closely the beam approximates a monoenergetic photon beam whose energy can be determined from the HVL. What is most commonly, although not always, available, is the kVp of the machine and the external or added filtration. All x-ray tubes have so-called inherent filtration, which is the window or port of the x-ray tube. In medical diagnostic units, the window or beam port through which the useful beam emerges is purposely made very thin, typically equivalent to 0.5 mm Al in attenuation, and hence provides little beam hardening.

Although the benefits of filtration with respect to improved radiographic images were known and understood as early as March 1896, within months of the discovery of x-rays (Magie 1896), initially diagnostic radiographs were made with no added filtration. Recommendations, albeit not specific as to thickness, were put forth in 1937 by the International Committee for Radiological Units which specified aluminum filters for x-rays of 20 to 120 kVp which incorporated the diagnostic x-ray energy range (ICRU 1937). This was consistent with, although not as specific as, the 1936 recommendation of the U.S. Advisory Committee on X-Ray and Radium Protection, the forerunner of the NCRP, which called for total filtration 0.5 mm of Al equivalent for radiographic installations, and 1 mm Al for fluoroscopy (NBS 1936). Generally, manufacturers of radiographic x-ray tubes complied 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, pp. 308-310).

Typical external or added filtration in the 1940's ranged from none to 1 mm Al. In 1949, the NCRP recommended 1 mm of added Al filtration for radiographing thick parts of the body such as the chest in 1949 and this thickness was in use during World War II in 100 mA units in larger military hospitals, and hence presumably at the various Manhattan District sites which were under the aegis of the U.S. Army (Olson, Trask and Dessent 1966). Subsequently, recommended thicknesses were increased not only for patient protection but for improved radiographic image quality; in 1955, the NCRP recommendation for diagnostic x-ray units called for 2 mm total Al filtration for new machines (NBS 1955), the filtration increased again to 2.5 mm by the 1960's for medical diagnostic units operating above 70 kVp (NCRP 1968). For machines already in operation, these recommended filter thickness might not have been utilized for some time after the date of the recommendation.

The relationship of beam intensity[1] to applied kVp and to filtration is complex and to some extent is machine specific and hence is best determined empirically. However, in the absence of empirical data for a specific machine, adequate contemporary empirical and theoretical data exist upon which to determine within a reasonable degree of uncertainty, the machine output. Additional filtration reduces the entrance skin exposure[2] (ESE), generally in an exponential manner. 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 per cent 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:

$$I = Io e^{-0.5t}$$

or

$$\ln (I/Io) = - 0.5\, t$$

in which t is the thickness of Al, in mm, and I and Io are the beam intensities with and without the filter, respectively. In the absence of specific measurements or empirical data, this correction can be applied to determine the effect of filtration on beam intensity, and is consistent with the guidance put forth in OCAS-IG-001 Revision 1 (2002).

Similarly, increasing the kVp will increase the beam intensity or exposure rate. This can be calculated using Kramer's rule, but such calculations are difficult, complex and time consuming, even with high speed computers, and are at best approximations. However numerous empirical studies of beam intensity as a function of kVp have been carried out over the years and provide ample credible evidence to show that for a given amount of filtration, increasing the applied kVp will increase the beam intensity according to the 1.7 power of the applied kilovoltage (Handloser 1951; Trout et al. 1952; Kathren 1965; Cameron 1970). In the absence of specific measurements or empirical data, this function can be applied to determine the effect of applied kilovoltage on beam intensity, and is fully consistent with the OCAS guidance document (OCAS 2002).

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[1] As used herein, beam intensity refers to the output of the machine in terms of exposure in the special sense per mAs. Exposure in the special sense is referenced to ionization in air and as such is not a dose quantity.

[2] Throughout this document, italics will be used to differentiate exposure in the special sense from exposure in the general sense. Thus *exposure* refers to exposure in the special sense. A brief discussion of exposure in both the general and special sense can be found in numerous publications, including NCRP Report 82 (1985) and ICRU Report 60 (1998). It is important to note that the definition and application of the quantity exposure and its concomitant unit the roentgen have undergone several important modifications over the years, which have been documented throughout the literature.

It should be noted that the effects of filtration and kVp tend to offset one another; addition of filtration reduces the exposure or dose per mAs, while increasing the kVp increases the exposure and dose per mAs. 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 dose from using increased kVp. However, there is not a direct correspondence or proportionality between the effects of filtration and kVp.

2.2 CURRENT AND EXPOSURE TIME

Diagnostic x-ray exposures are typically specified in terms of milliampere-seconds (mAs), the product of x-ray tube current and the exposure time. Other factors being equal (e.g. kVp, filtration, film, development and screen combination) radiation exposure is thus proportional to the number of mAs. The current in an x-ray tube refers to the number of electrons accelerated across the evacuated volume of the x-ray tube, flowing from the cathode to the anode. For a given applied kilovoltage, the number of x-ray photons produced, and hence the exposure, will at least in theory be directly proportional to the x-ray tube current, and indeed this is and has been historically true for most medical radiography units over their designed tube current range. Thus, in the absence of measurements or other data or information to the contrary, it is reasonable and consistent with long standing radiographic practice (Sante 1946) to assume linearity of beam intensity and hence patient dose with tube current.

Exposure time refers to the time that the beam was on or the machine was producing x-rays and is, for all practical purposes, linear with exposure. To avoid or minimize image blurring from the beating heart, exposure time was minimized, and the current concomitantly and proportionately increased to obtain the desired exposure in terms of mAs. However, from a dose reconstruction standpoint, it should be noted that earlier medical radiographic units were equipped with mechanical timers whose accuracy was not as good as the electronic timers used on later model apparatus. Gross bias errors in timer accuracy are unlikely in that these would result in over- or underexposure of the radiograph and so would be quickly detected and corrected. More subtle are small random errors, which might produce uncertainties of perhaps + 20 per cent in the exposure.

Chest photofluorography, which resulted in very much greater patient doses from a diagnostic procedure, was used sporadically until as late as the early 1960's. Photofluorography used a smaller film (4 x 5 inches), a smaller SSD (42 inches), and both a higher kVp and typically a several fold greater exposure in terms of mAs. Exposure was regulated by photometers, which utilized the exposure to the film to determine the time of exposure.

2.3 DISTANCE

X-ray beam intensity is a function of distance from the target, approximating inverse square at large distances from the tube. Radiographic chest films were taken at a standard source to image distance (SID) of 72 inches; the source refers to the focal spot of the tube and the image to the plane of the film. The distance to the patient, sometimes expressed in terms of the source to skin distance (SSD), is somewhat smaller since the patient is positioned between the source and the film cassette and hence, the entrance skin exposure (ESE) to the patient is somewhat greater than the exposure at the plane of the film. In addition, patient attenuation would further reduce or attenuate the number of photons reaching the film. To compensate for the increased attenuation provided by a larger patient, x-ray technicians would sometimes increase the beam settings for a large patient, or, if the machine was so equipped, might use a high speed Bucky diaphragm, likely with a somewhat higher kVp. It thus may be appropriate for an individual dose reconstruction to increase the ESE or skin entrance kerma for a large or stout patient. Based on standard contemporary techniques (Picker 1941; Fuchs

1958; Cahoon 1961) for patients with a chest thickness of 25-27 cm, an increase of +50 % from the ESE to the average patient should be sufficiently conservative; for still larger patients, a factor of 2 would be appropriate. The average worker chest size is taken to be 22-24 centimeters.

2.4 COLLIMATION AND WAVEFORM CHARACTERISTICS

Among the other factors that potentially affect patient dose are collimation and waveform. X-ray waveforms are of three types: half wave rectified, which is almost never seen; full wave rectified, which is typical of virtually all medical radiographic units, and constant potential. A half wave rectified machine produces 60 half sinusoidal shape pulses of x-rays per second, each with a duration of 1/120 of a second. A full wave rectified machine produces 120 half sinusoidal pulses of x-rays per second, each with a duration of 1/120 second. Thus, for a given setting of kVp and mA, the intensity of the beam from a half wave rectified machine will be half that of the beam from the full wave rectified type. A constant potential machine produces a more or less steady (i.e. unpulsed) output of x-rays and has a somewhat greater beam intensity – approximately 10 per cent – as compared with a full wave rectified machine operating at the same kVp and mA.

Collimation refers to the size of beam. Early, the philosophy was to use a fairly large aperture (i.e. limited collimation) to ensure that the entire area of interest was included in the radiograph. Subsequently, because of patient protection concerns, beams were collimated such that the smallest beam consistent with the area of interest was used, thereby limiting the area of the patient exposed, and, in the case of chest radiography, minimizing dose to organs such as gonads, thyroid, and gastrointestinal tract. A practical check of collimation can be made by reference to the radiograph; a well collimated beam will leave a small unexposed area or penumbra effect at the edges of the radiograph, while a poorly collimated beam will produce a radiograph that is exposed all over its area. Beam diameter limiting cones were used in radiography during the 1940's and beyond to improve radiographic image quality by reducing scatter (Glasser et al. 1944, p. 136) and were sometimes equipped with an Al filter, 1 mm in early years and thicker later on.

Due to the reported variation in the literature and measurement data on the effects of collimation, the claimant favorable assumption of no external collimation of the primary beam should be used when measurement data, technique, or other information to describe the collimation are not available for x-rays taken prior to 1970. This is based on the following claimant favorable assumptions and professional judgment:

1. In the late 1950s, there was significant research into the gonadal dose and the reasons for the observed variation in dose. This research described the effects of filtration, collimation, and centering. By the early 1960s, techniques were being modified incorporating additional collimation. While these techniques were likely fully incorporated at most DOE facilities by 1965, to allow for the possibility that some smaller facilities might not have had the resources to update their equipment and to be claimant favorable, the year 1970 was selected.

2. In 1968, the National Council on Radiation Protection and Measurements (NCRP) in Report 33 updated their guidance on medical x-ray protection. While many DOE facilities had probably already incorporated the guidance in this report, some smaller facilities might not have incorporated the guidance by 1968. To ensure that these facilities were in fact in conformance with the 1968 recommendations, an additional two-year period was added.

3. By the late 1950s, reports in the literature of most of the surveys of medical x-ray facilities revealed low gonadal doses, indicating adequate collimation. A few surveys clearly indicated the use of collimation was limited. Of the eight surveyed facilities at Oak Ridge, only one

(13%) had a moderately high male gonadal dose (5 mrad). All of the other facilities, the male gonadal dose was less than 2 mrad. Variation between the other facilities appeared to be the result of differences in the use of filtration and cone size. Since most facilities were using some form of collimation by the late 1950s, by the mid 1960s most, if not all, facilities were probably using some form of collimation. Since references as to when all facilities were using adequate collimation were not found, professional judgment was used to estimate this time period to be the mid 1960s. To fully assure claimant favorability, this assumption has been further expanded by 5 years to 1970 to allow for the uncertainty in professional judgment.

The organ doses in this document apply to chest x-rays performed after 1970 which are assumed to be properly collimated. Organ doses for chest x-rays performed before 1970 which may have been minimally collimated, will be published in a separate document.

2.5 SCREENS, GRIDS, AND OTHER FACTORS POTENTIALLY AFFECTING PATIENT DOSE

A number of other factors also affect the x-ray exposure required to obtain a proper radiograph and hence the dose to the patient. However, knowledge of these factors is unnecessary for dose reconstruction purposes if beam measurements are available or if the primary machine characteristics of applied kilovoltage (kVp), time and current (mA) are known along with the amount of primary beam filtration, although they can be used as additional confirmation of the applicability of the reconstructed dose. Hence, for completeness, only brief mention will be made of these factors, which are: tube housing, type and speed of the film, development procedure, screens, and grids.

X-ray tubes used for diagnostic radiography are typically enclosed in a protective lead tube housings with the primary beam brought out through a port or window in the side of the housing. Although some reduction of the dose to the patient is achieved, largely through elimination of scattered radiation and improved collimation, this so-called diagnostic tube housing is primarily for the purpose of protection of the operator and unexposed x-ray film and nearby individuals other than the patient. The issue is moot, however, since virtually all x-ray tubes used to x-ray the DOE weapons worker cohort have been equipped with protective tube housings, which limited leakage to < 0.1 R/hr at one meter from the tube.

The exposure needed for a suitable diagnostic radiograph is in some measure a function of film speed and development. So called fine grain emulsions produce a superior radiographic image but require additional exposure as compared with so called fast films, which typically have a larger grain size. Underdevelopment of films will also require additional exposure to achieve satisfactory radiographic quality. Intensifying screens are used within the cassette to intensify the radiographic effect and thereby effectively increase film speed and reduce patient dose. Film speeds have typically increased since the 1940's and reduced patient doses appreciably, perhaps by half. Grids, specifically the Potter Bucky diaphragm (also more or less colloquially known as a Bucky) are sometimes utilized for thick section radiography, but rarely used for chest radiography except with very large patients. In any case, the above are all factored into the technique (i.e. kVp, mA) that is used and except in rare instances and a virtually complete absence of other data, are not of importance in dose reconstruction.

2.6 SUMMARY AND APPLICATION OF TECHNICAL FACTORS

For convenience and possible application to cases in which other and more suitable data are not available, or for generic use, the effect of various technical factors has been tabulated below in Table 2.6-1. Samples of technique factors are represented in Table 2.6-2.

Table 2.6-1. Relationship of Beam Intensity and Various Technical Factors

Parameter	**Units**	**Relationship with intensity**
Applied voltage	kVp	Intensity proportional to 1.7 power of kVp
Tube current	mA	Linear
Exposure time	s	Linear
Filtration	mm Al	Intensity decreases by ~40% for each additional mm Al
Patient Size (chest thickness)	25-27 mm > 27 mm	Dose increased by factor of 1.5 Dose increased by factor of 2
Distance	d	Approximately inverse square relationship (1/d2) holds for distances > about 30 cm from target
Uncertainty	± 30%	Assumes all errors are positive, + 30% should be used

Table 2.6-2. Samples of Technique factors used for different types of X-ray equipment.

Machine	View	Current (mA)	Voltage (kVp)	Exposure time (sec.)
Type II	PAa	300	110-120	1/30
Type III	PA	300	120	1/40
Type IV	PA	300	120	1/40

[a] PA indicates a posterior/anterior view, the average PA chest measures 26 cm. The average Lat. chest measures 34 cm.

3.0 Reconstruction of Diagnostic Medical X-Ray Doses

Not all workers were required to undergo medical diagnostic x-ray examinations as a condition of employment, and among those who were, the procedure was usually limited to a single PA chest film, although a lateral chest film might also be taken. Some workers were examined with chest photofluorographic units which produced a much greater dose than the standard PA radiograph. A very small fraction might have undergone lumbar spine or other specific procedures if there was a medical indication.

The incidence of defective films necessitating retakes is not known, but it is likely to have been very small, and certainly no more a few per cent and probably much less. Trout et al (1973) in their analysis of the rejection rate of chest radiographs obtained during the Coal Mine “Black Lung” program reported an average rejection rate of 3% among 67,000 radiographs. Retakes should serve as a signal to give special consideration to the evaluation of technique factors, and hence the resultant dose calculations. A retake in a very large individual might serve as a signal that the initial radiograph was taken with technique factor settings suitable for a smaller person, and that the second radiograph reflected an additional and larger dose. Retakes in African-Americans may signal that the initial exposure was too great, and indicative of an overriding of the standard or automatic technique factors because of a perception held by some x-ray technicians that African-Americans had greater bone density or other characteristics that required additional exposure. Similarly, retakes in females may be indicative of manually altered settings to increase beam intensity under the misimpression that additional exposure was required for women because of the larger amount of breast tissue. Unless machine settings have been recorded, it is impossible to determine whether the retake was necessitated by an arbitrary manual increase in machine settings to obtain a greater exposure. If machine settings are available, then an adjustment for the increased doses can be made using the data provided in Table 2.6-1. If machine settings are not available, then to ensure claimant favorability for African-Americans and women whose records indicate retakes, an upward adjustment of the organ doses from the first radiograph is indicated; increasing doses by a factor of two in these cases should more than compensate for the supposed additional exposure.

Diagnostic medical x-ray dose reconstruction is best accomplished when actual measurements of beam intensity are available. Use of actual measurement data is the simplest and most direct means of assessing diagnostic medical x-ray doses, typically requires few, if any assumptions, and has the least amount of uncertainty. Hence, use of actual measurement data, where available, is preferred for diagnostic medical x-ray dose reconstruction and should be used if available. Actual beam measurements are most likely to provide the most accurate estimates of organ doses.

X-ray output measurements are likely to be unavailable, particularly prior to about 1980. In the absence of suitable measurement data, medical diagnostic x-ray dose reconstruction can be accomplished using technique factors along with published output data that provide beam intensity per mAs as a function of kVp, filtration, and distance. The use of technique factors will typically require a number of assumptions, and these, of course, should be claimant favorable. If both measurement data and technique factors are unavailable or unknown, then dose estimates can be

made using the default values shown in Table 3.3-1. Use of default values is a last resort. Reiterating, the first choice should be to use actual beam measurement data when available.

3.1 RECONSTRUCTION WHEN MEASUREMENTS ARE AVAILABLE

Although beam output measurements may typically be unavailable, diagnostic medical x-ray dose reconstruction using actual measurement data is the preferred method for determining the dose to the worker from this source, so much so that special effort to determine if such measurements have been made is justifiable. Beam output measurements are typically made in terms of exposure and are quantified in units of R, and depending on the measurement device and technique may have a wide range of uncertainty. The best measurements are those made with integrating ionization chambers designed for medical x-ray applications; until about 1970 or so, there were two such instruments in common usage and availability in the United States, the Victoreen R-meter, or Landsverk L series ion chambers. More recently, a wide variety of such instruments have become available. Measurements with R-meters and similar chambers, if properly done, have a high degree of reliability and a low degree of uncertainty. Generally, the uncertainty of properly done measurements in the energy region of interest should not exceed + 2 % of the measured value (Kathren and Larson 1969).

Other integrating devices such as film, TLD and pocket ionization chambers have also been used for beam output measurements. Measurements made with these types of dosimeters should be used with great caution. All are, to varying degrees, energy dependent, and correction for beam energy is a necessity. This requires knowledge of the x-ray beam energy and the response of the dosimeter as a function of energy. Typically, pocket ionization chambers provide the least reliable measurements with the greatest uncertainty, and results obtained with these devices are highly suspect and should be used with great caution. Film and TLD, if appropriately calibrated to the beam energy, can provide satisfactory measurements, albeit with a considerable degree of uncertainty. The widely used LiF TLDs, compared with higher Z phosphors and film, show relatively good energy and other response characteristics, and, if properly used can provide uncertainties similar to those of R-meters. Considerably greater uncertainty – perhaps on the order of several tens of a percent – may apply to film dosimeters. No specific uncertainty values can be provided here, as each film dosimeter and TLD system is different, and reference to the literature is necessary to determine the appropriate values for specific dosimeter systems.

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 2.6-1. The ESE will, of course be in units of R, which must be converted to kerma and then to organ dose. As discussed above, an exposure of 1R is typically taken to be equal to a kerma of 1 rad (10 mGy); actually, 1R is slightly less than 1 rad (10 mGy) of kerma, but the difference is small and making the numerical equivalence greatly simplifies the dose reconstruction as well as providing a small additional measure of claimant favorability.

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 through A9 of ICRP Publication 34 (ICRP 1982). Use of these tables requires knowledge of the x-ray beam quality expressed in terms of the HVL in Al. If the kVp and filtration are known, HVLs can be estimated from the data given in Table A16 of ICRP Publication 34 (ICRP 1982, p. 77) or Table B.2 in NCRP Report No. 102 (p. 98). In general, the greater the kVp and filtration, the greater the HVL. If the actual beam quality is unknown, as is likely the case, to ensure claimant favorability a higher rather a lower HVL should be assumed. In the absence of actual data, recommended default values for beam quality are 2.5 mm Al HVL for radiographs taken prior to 1980, and 4.0 mm for radiographs subsequently. These values are likely overestimates of HVL and hence are claimant favorable.

However, Tables A2 to A9 in ICRP Publication 34 do not include all the organs that have been identified in the IREP code. For those organs included in the IREP but not specifically identified in ICRP Publication 34, use of the dose conversion coefficient for the organ specified in ICRP Publication 34 that is anatomically the closest would seem to be a reasonable and simple first order approach that generally would be claimant favorable or neutral. Thus, the factor for lung would be applied to all other organs within the thoracic cavity – i.e. thymus, esophagus, and stomach. Since an appreciable fraction of the skeleton, and in particular the trabecular bone which has a large surface to volume ratio and the sternum lies within the trunk, the factor for lung would also be applied to the bone surfaces. For organs in the abdomen – i.e. liver, urinary bladder, and colon – the dose conversion coefficient for ovary would be used. For the eye, the analogous organ is the thyroid. Skin dose can be obtained by reference to Table B.8 in NCRP Report No. 102 (NCRP 1989, p. 103), which provides backscatter factors for different beam qualities and field sizes. For chest radiography, a backscatter factor of 1.35 is recommended to ensure claimant favorability.

It is useful to prepare a summary table of beam parameters as shown in Table 3.1-1. This table is taken from actual data and measurements available for the Hanford Site and is shown here as an example of what a summary table should include where measurements are available. The table includes not only the measured values for given time periods, but a reference to those values as well as other salient data pertaining to beam and exposure.

Table 3.1-1. Example of Summary Data Based on Actual Beam Measurements for the Hanford Site.

Date Measured	10/18/1999	2/04/98	4/22/1997	11/11/1993	3/30/1990	1/21/1988
Procedure	Chest PA 14"×17"	Chest PA 14"×17"	Chest PA 14"×17"	Chest PA 14"×17"	Chest PA 14"×17"	Chest PA 14"×17"
Machine type	XMA - 360	XMA - 360	CONXI Type 12	CONX Type 12	CONX Type 12	CONX Type 12
Machine settings kVp:	110	110	110	110	110	110
mA	300	300	300	200	200	200
Exposure time	1/60 sec	1/60 sec	1/30 sec	1/30 sec	1/30 sec	1/20 sec
mAs	5	5	10	6.7	6.7	10
Added filter	2.7 mm	2.7 mm	2.5 mm	2.5 mm	2.5 mm	2.5 mm
Filtration used for calcs.	4.0 mm	4.0 mm	4.0 mm	4.0 mm	4.0 mm	4.0 mm
Source to skin distance	72"	72"	72 "	72 "	72 "	72 "
Entrance skin exposure	11 mR	11 mR	17 mR	21mR	21 mR (Assumed)	35 mR
mR/mAs	2.2	2.2	1.7	3.3	3.3	3.5

Date Measured	10/18/1999	2/04/98	4/22/1997	11/11/1993	3/30/1990	1/21/1988
Date range		2/98 to date	4/97 to 2/98		3/90 to 4/97	
Reference	Washington State Dept. of Health Measurement	Washington State Dept. of Health Measurement	Washington State Dept. of Health Measurement	Washington State Dept. of Health Measurement	Measured at 11.7 mR by State. The 1993 value was used, as it was higher for same settings& machine.	Washington State Dept. Of Health Measurement

Date Measured	**1/20/1988**	**1/28/1983**	**4/12/1959**	**2/1/1946**	**Before 2/46**
Procedure	Chest PA 14"×17"	Chest PA 14"×17"	Chest PA 14"×17"	Chest PA 14"×17"	Chest PA 14"×17"
Machine type	CONX Type 12	G.E. DXR 750	Unknown	Unknown	Unknown
Machine settings kVp:	110	100	80	80	Unknown
mA	100	200	300	500	Unknown
Exposure time	1/10 sec	1/20 sec	1/30 sec	1/20 sec	Unknown
mAs	10	10	10	25	Unknown
Added filter	2.5 mm	2.5 mm	1.5 mm	1.5 mm	1.5 mm
Filtration used for calcs.	4.0 mm	2.5 mm	2.5 mm	2.5 mm	2.5 mm
Source to skin distance	72 "	72 "	72 "	72 "	72 "
Entrance skin exposure	35 mR	35 mR	40 mR	79 mR	120 mR
mR/mAs	3.5	3.5	4	3.2	Unknown
Date range		1/83 to 3/90	4/59 to 1/83	2/46 to 4/59	
Reference	Washington State Dept. of Health Measurement	Kathren to Heid memorandum Dated 1/28/83	Rising & Soldat letter to Norwood dated 4/30/59	Mancuso et al. Dated 1966	Based on experience & references of early 1940s x-ray dose. Assumed for Hanford.

3.2 RECONSTRUCTION USING TECHNIQUE FACTORS

When beam measurement data are unavailable, as is likely to be the case, technique factors can be used to obtain reasonable estimates of exposure. The basic data required are kVp, filtration, exposure in mAs, and distance. Beam output data are available from a number of publications, including NCRP Report No. 102 (NCRP 1989). Table B.3 in this report (p. 99) provides average air kerma rates for medical diagnostic x-ray equipment operating at various kVps with 2.5 mm Al filtration at distances from 30 to 182 cm from the source. Correction for different thickness of Al filtration can be made by reference to Table 2.6-1. Alternatively, Figure B.1 (p. 109) in NCRP Report No. 102 provides a graphical representation of air kerma at 100 cm for various values of kVp and filter thickness > 2.5 mm Al. Using these tables, a reasonable estimate of beam output and hence entrance kerma can be obtained. Once the entrance kerma has been determined, organ doses are determined in the manner described above for reconstruction using measurement data.

3.3 RECONSTRUCTION USING DEFAULT VALUES

Default values of entrance kerma have been developed for the three most commonly used occupational medical diagnostic x-ray procedures: PA chest radiography; lateral chest radiography; photofluorographic chest films when actual measurement data or knowledge of technique factors are absent. The default values are considered to be maxima developed from review of patient doses as reported in the literature, machine characteristics, and knowledge of x-ray procedures used during the time periods indicated. Sufficient conservatism was included in the determination of the default values to ensure with near certainty (99+ per cent confidence) that the actual exposures from the specified procedures would not exceed the default values, thus ensuring claimant favorability. In determining these factors, it was assumed that a minimum of filtration was used along with low kilovoltage techniques, slow film speeds with standard development, and no additional collimation or use of cones. The default entrance kerma values for the three procedures are given in Table 3.3-1.

Table 3.3-1. Default Dose Values by Procedure

Period	Entrance kerma, cGy PA chest	Entrance kerma, cGy Lateral chest	Entrance kerma, cGy Photofluorographic chest
Pre-1970	0.20	0.50	3.0
1970-1985	0.10	0.25	
Post 1985	0.05	0.13	

The above default values can then be used as described above in lieu of actual measurement data or entrance kerma derived from technique factors.

4.0 Application and Reporting of Occupational Medical X-Ray Dose Reconstruction

Table 4.0-1 provides organ dose conversion factors and organ dose calculations for dose reconstruction for chest x-rays performed after 1970.

Table 4.0-1. Organ doses for Default Entrance Kerma Values

Organ	View	Dose Conversion Factor (mGy per Gy air kerma) [(a)] HVL 2.5 mm Al	Organ Dose 1970-1985 (rem) (c,d)
Thyroid	PA	32	3.20E-3
	Lat	115	2.88E-2
Eye/Brain	PA	32	3.20E-3
	Lat	115	2.88E-2
Ovaries	PA	1	1.00E-4
	Lat	0. 6	1.50E-4
Liver/Gall Bladder	PA	1	1.00E-4
	Lat	0. 6	1.50E-4
Urinary Bladder	PA	1	1.00E-4
	Lat	0. 6	1.50E-4
Colon Rectum	PA	1	1.00E-4
	Lat	0. 6	1.50E-4
Testes	PA	0.01	1.00E-6
	Lat	0.1	2.50E-5
Lungs (male)	PA	419	4.19E-2
	Lat	193	4.83E-2
Lungs (female)	PA	451	4.51E-2
	Lat	220	5.50E-2
Thymus	PA	451	4.51E-2
	Lat	220	5.50E-2
Esophagus	PA	451	4.51E-2
	Lat	220	5.50E-2
Stomach	PA	451	4.51E-2
	Lat	220	5.50E-2
Bone Surfaces	PA	451	4.51E-2
	Lat	220	5.50E-2
Remainder	PA	451	4.51E-2
	Lat	220	5.50E-2
Breast	PA	49	4.90E-3
	Lat	255	6.38E-2
Uterus	PA	1.3	1.30E-4
	Lat	0.6	1.50E-4
Bone Marrow (male)	PA	92	9.20E-3
	Lat	37	9.25E-3
Bone Marrow (female)	PA	86	8.60E-3
	Lat	29	7.25E-3
Total Body (male)	PA	131	1.31E-2
	Lat (b)	64	1.60E-2
Total Body (female)	PA	118	1.18E-2
	Lat (b)	60	1.50E-2
Skin (e)	PA		1.35E-1
	Lat		3.38E-1

Table 4.0-1. Organ doses for Default Entrance Kerma Values (cont'd)

Organ	**View**	**Dose Conversion Factor (mGy per Gy air kerma) HVL 4.0 mm Al [(a)]**	**Organ Dose Post 1985 (rem) (c,d)**
Thyroid	PA	78	3.90E-3
	Lat	164	2.13E-2
Eye/Brain	PA	78	3.90E-3
	Lat	164	2.13E-2
Ovaries	PA	5.2	2.60E-4
	Lat	2.5	3.25E-4
Liver/Gall Bladder	PA	5.2	2.60E-4
	Lat	2.5	3.25E-4
Urinary Bladder	PA	5.2	2.60E-4
	Lat	2.5	3.25E-4
Colon/Rectum	PA	5.2	2.60E-4
	Lat	2.5	3.25E-4
Testes	PA	0.01	5.00E-7
	Lat	0.1	1.30E-5
Lungs (male)	PA	628	3.14E-2
	Lat	313	4.07E-2
Lungs (female)	PA	674	3.37E-2
	Lat	351	4.56E-2
Thymus	PA	674	3.37E-2
	Lat	351	4.56E-2
Esophagus	PA	674	3.37E-2
	Lat	351	4.56E-2
Stomach	PA	674	3.37E-2
	Lat	351	4.56E-2
Bone Surfaces	PA	674	3.37E-2
	Lat	351	4.56E-2
Remainder	PA	674	3.37E-2
	Lat	351	4.56E-2
Breast	PA	116	5.80E-3
	Lat	343	4.46E-2
Uterus	PA	5.2	2.60E-4
	Lat	2.1	2.73E-4
Bone Marrow (male)	PA	178	8.90E-3
	Lat	76	9.88E-3
Bone Marrow (female)	PA	172	8.60E-3
	Lat	59	7.67E-3
Total Body (male)	PA	192	9.60E-3
	Lat	106	1.38E-2
Total Body (female)	PA	178	8.90E-3
	Lat	99	1.29E-2
Skin (f)	PA		7.00E-2
	Lat		1.82E-1

a. Dose conversion Factors from Tables A.2 through A.9, ICRP Publication 34 (1982).

b. The values for lateral x-rays from ICRP 34 (1982) appear to be switched in Table A9. All the other values for male to female dose have the male values higher than the female. The factors were changed to agree with the other factors listed for other HVLs.

c. Source to Image-Distance 183 cm.

d. Image Receptor Size (cm) 35.6 x 43.2.

e. Calculated using backscatter factor of 1.35 from NCRP Report 102, Table B-3.

f. Calculated using backscatter factor of 1.35 from NCRP Report 102, Table B-3.

5.0 Uncertainty Analysis for Diagnostic Medical x-Ray Doses

Error (deviation from the correct, true, or conventionally accepted value of a quantity) and uncertainty (defined in terms of the potential range of a stated, measured, assumed, or otherwise determined value of a quantity) provide an indication of the confidence or validity of the dose estimates. 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, which in turn is expressed as a percent. Thus, the 99% confidence level indicates that the correct or true value, although not actually known, has a 99% probability of falling within the range cited. The statement of confidence level 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 factors can introduce uncertainties or affect the X-ray machine output intensity and dose to the worker. However in practice only five factors can be reasonably considered to have a meaningful or significant impact on dose uncertainty. These are:

1. Measurement error
2. Variation in applied kilovoltage
3. Variation in beam current
4. Variation in exposure time
5. Distance from the worker to the source of the X-rays (SSD)

The influence of such other factors as use of screens, grids, reciprocity failure, film speed, and development, while potentially variable, do not affect the beam output intensity per se except indirectly insofar as these may determine the exposure settings (i.e. kVp, mA, and time) used.

Medical x-ray doses, when measured, were largely derived from actual measurement of X-ray machine output with R-meters or similar ionization chamber devices suitably designed for measurement of photons in the medical x-ray energy range. If properly calibrated and used, 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 a somewhat smaller uncertainty, perhaps on the order of + 1% (NBS 1985, 1988), for conservatism, the uncertainty range of + 2% should be applied to measurements of X-ray intensity.

Theoretically, for a given set of machine settings and parameters, X-ray output should be constant and unvarying. However, this is not true in practice. Although output is essentially constant unless focal spot loading occurs, as might be the case when the power rating of the machine is exceeded, it is unlikely that power ratings were ever exceeded because such an event would be difficult to achieve in practice and could result in damage to the X-ray tube. However, even with the use of constant voltage transformers to control line voltages, slight variations might occur in line voltage input or other internal voltages, which in turn could alter the kVp of the output beam. In general, for a given kVp setting, variation in kVp falls within + 5% of the machine setting (Seibert et al. 1991). Since as noted above, beam intensity is approximately proportional to the 1.7 power of the applied kilovoltage; this

translates to an uncertainty of approximately ± 8.6% with respect to output beam intensity in the 80 to 100 kVp range used for diagnostic chest radiographs. For conservatism, this is rounded up to + 9%.

Similarly, slight variations in tube current are normal; as a tube ages, or heats up from use, current can change and typically will drop. With all other factors constant, beam intensity will be reduced in direct proportion to the change in tube current. Typically, the reduction in beam output from current variation is not more than a few percent under normal operating conditions; large decreases are readily detectable and manifest themselves as underexposed radiographs and result in maintenance on the machine to restore the output or, as a temporary measure, an increase in the current or kVp to provide the necessary intensity for proper radiography. For a given kVp setting, the output of the beam is a function of the tube current, which in turn is measured by a milliammeter, which measures average tube current. The measurement is subject to uncertainties; there might be minor changes in output as the tube heats from normal use. These variations are typically small, and the estimated uncertainty in beam intensity or output attributable to current variation is + 5%.

Another parameter that has potential to affect the dose from a diagnostic radiograph, perhaps significantly, is the time of exposure. The potential importance of this parameter is underscored by noting that virtually all medical diagnostic medical x-ray units used in the DOE complex were of the full wave rectified type. A full wave rectified machine produces 120 pulses of X-rays per second. Thus, in a typical radiographic exposure time of 1/20 of a second, only six pulses would result. A small error 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 of a second, the change in output corresponding to a deviation of + 1 pulse is + 25%. Early mechanical timers were notoriously inaccurate; accuracy improved significantly with the introduction of electronic timers. Measurements of reproducibility made in the late 1980s and beyond by the State of Washington for the machines at Hanford suggest that the timers, and indeed the entire X-ray output, were fairly constant. However, for conservatism, the assumed uncertainty in beam output attributable to timers has been taken to be + 25%.

The final factor likely to affect worker dose relates to distance from the source of the X-rays, which is an important determinant of the entrance skin exposure from which organ doses are calculationally derived. For a given individual, the SSD will be determined largely by the body thickness of the worker and the accuracy of the positioning. For a typical worker, the estimated variation in SSD is no more than a few centimeters, with an upper limit of perhaps 7.5 cm. Using inverse square, this indicates an uncertainty of + 10% from this source.

There are two approaches to determine the combined uncertainty from the five potential sources of dose uncertainty listed above. The first, and most conservative in that it gives the greatest range, would be to assume that the uncertainties are additive, which would give an uncertainty range of 2 + 9 +5 + 25 + 10 = + 51%. However, a more reasonable approach would be to assume that the uncertainties are in fact random, and to compute the statistical root mean square (RMS) value. The RMS value is simply the square root of the sum of the squares, and computes as + 28.9%. Rounding this up to + 30% would seem to provide an adequate and suitably conservative indication of uncertainty. Thus, for an individual ESE or derived organ dose, an uncertainty of + 30% at the 99% confidence level can be assumed; for further conservatism it might be appropriate to assume that errors are all positive, and only + 30% should be used.

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