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Circulation. 2007;116:1290-1305
doi: 10.1161/CIRCULATIONAHA.107.688101
Circulation is published by the American Heart Association, 7272 Greenville Avenue, Dallas, TX 75231

Print ISSN: 0009-7322. Online ISSN: 1524-4539

Radiation Dose to Patients From Cardiac Diagnostic Imaging

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The volume of cardiac diagnostic procedures involving the use of ionizing radiation has increased rapidly in recent years. Whereas in 1990, fewer than 3 million nuclear cardiology studies were performed in the United States, by 2002 this figure more than tripled to 9.9 million.[1] Cardiac computed tomographic (CT) volume doubled between 2002 and 2003, to 485 000 cases,[2] and has continued to grow since then. The volume of procedures performed in cardiac catheterization labs increased from 2.45 million in 1993 to 3.85 million in 2002.[3]

The powerful diagnostic and risk-stratification data provided by these procedures play a central role in clinical cardiology and have contributed to the decrease in morbidity and mortality from coronary heart disease. Nevertheless, performance of any diagnostic test requires a careful assessment of the risks and benefits of the test and optimization of protocols to minimize risks to patients, staff members, and the public. Procedures that utilize ionizing radiation should be performed in accordance with the As Low As Reasonably Achievable (ALARA) philosophy. Thus, physicians ordering and performing cardiac imaging should be very familiar with the dosage of radiation from cardiac diagnostic tests and ways in which dose can be minimized. In this report we discuss the measurement of radiation and the dosimetry of commonly performed cardiac diagnostic imaging tests, including nuclear scintigraphy, CT for calcium scoring and coronary angiography (CTCA), and conventional coronary angiography (CCA). For each modality, we address the terminology and methodology used to quantify radiation received by patients, doses to patients with typical protocols, and dose-reduction techniques.

General Terminology Used in Quantifying Radiation

Biological effects of ionizing radiation can be classified as deterministic or stochastic. Deterministic effects such as skin injuries and cataract formation occur predictably when dose exceeds a certain threshold, whereas stochastic effects such as cancer incidence and germ cell mutations occur with a probability that increases with dose.

Numerous quantities and units are used to measure radiation, some of which are summarized in Table I in the online-only Data Supplement.[4] Some ambiguity exists in the terminology used in the literature, confounded by multiple sets of units and changing nomenclature between guidelines. This nomenclature includes both general terms to describe quantities of radiation and specific terminology applicable to particular types of radiation sources or imaging modalities.

Organizations Setting General Terminology

The currently used general terminology is a product of the effort of multiple international organizations, notably the International Commission on Radiation Units and Measurements (ICRU), International Commission on Radiological Protection (ICRP), and Conférence Générale des Poids et Mesures (CGPM; General Conference on Weights and Measures). The ICRU, initially known as the International X-ray Unit Committee, was founded in 1925 by the International Congress of Radiology. Its principal objective is to develop international recommendations on quantities and units of radiation, on procedures for the measurement and application of these quantities, and on physical data required for the application of these procedures. The ICRU is composed of a chair and 13 commission members who are all physicists or physicians, assisted by 20 report committees addressing specific topics; to date, 76 reports have been issued. The ICRP, founded in 1928 as the International X-ray and Radium Protection Committee, is a daughter organization of the International Society of Radiology, although its work now focuses on all aspects of protection from ionizing radiation, not limited to medical applications. It is composed of a main commission, with a chair and 12 members with backgrounds in medicine, physical and biological science, engineering, and epidemiology, and 5 standing committees focusing on different aspects of radiological protection, supported by a scientific secretariat. Toward its goals, it has issued ≈100 reports

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The online-only Data Supplement, consisting of tables, is available with this article at http://circ.ahajournals.org/cgi/content/full/116/11/1290/DC1.

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(*Circulation.* 2007;116:1290-1305.)

***Circulation* is available at http://circ.ahajournals.org** **DOI: 10.1161/CIRCULATIONAHA.107.688101**

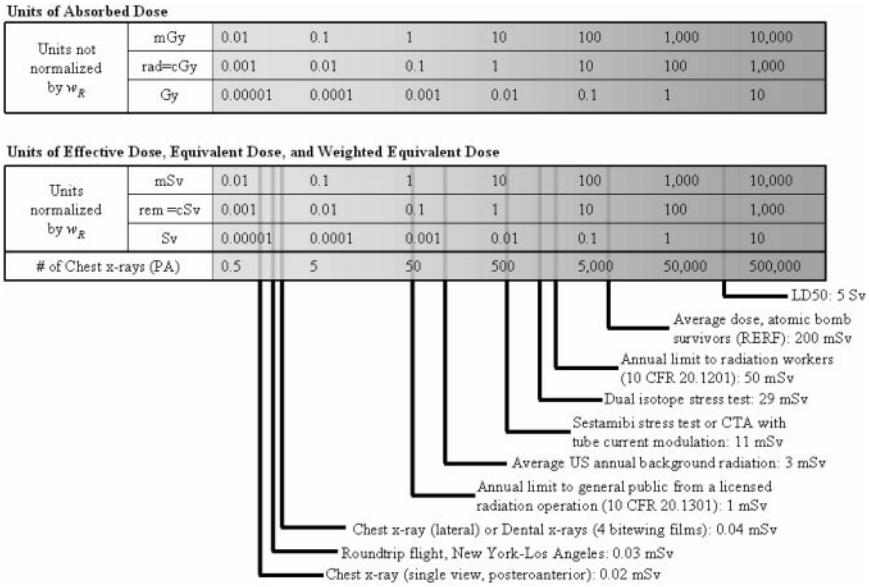

Figure 1. Top, Relationship between units of organ absorbed dose, using a log scale. Bottom, Relationship between units of effective dose, with effective doses of some representative radiation sources. CFR indicates Code of Federal Regulations; RERF, Radiation Effects Research Foundation; LD50, lethal dose to 50% of individuals; and w_R, radiation weighting factor (=1 for x-rays and γ-rays). Chest x-ray dose of 0.02 mSv per European Commission.[7]

authored by expert panels providing specific recommendations in 1 area of radiological protection, as well as periodically updated general recommendations, reflecting the state of knowledge on the biological effects of ionizing radiation. The CGPM is 1 of 3 linked organizations established by the Convention du Mètre, an international treaty signed in 1875 and now with 51 states as members, that have authority to conduct international activities in standardizing measurement. The CGPM established the Système International d'Unités (SI; International System of Units) in 1960 and now meets every 4 years to maintain and update it. Delegates to the CGPM are typically representatives of national standards or metrology institutes, although other related international organizations such as the International Atomic Energy Agency are represented as well. Although the CGPM has attempted to keep the SI as parsimonious as possible, special units have been introduced to quantify ionizing radiation to avoid its underestimation and thereby safeguard human health.[5]

Nomenclature

While the term *exposure* is used in a general sense to apply to an occurrence in which an individual is exposed to radiation, it also has a specific technical definition. Exposure equals the total charge of ions of 1 sign (positive or negative) produced per unit of dry air by a given amount of ionizing radiation. In SI units, exposure is measured in terms of coulombs (C) per kilogram. Exposure is also commonly measured in units of roentgens (R), where 1 R=2.58×10^{-4} C/kg. A related quantity is air kerma. Kerma, an acronym for "kinetic energy released in material,"[6] is the sum of the kinetic energy of all of the charged particles liberated per unit mass of a material by an amount of ionizing radiation. When that material is air, the kerma is referred to as air kerma. Thus, whereas exposure measures electric charge produced in air per unit mass from an amount of ionizing radiation, air kerma measures its energy produced in air per unit mass. While often easy to measure, exposure and air kerma specifically measure ionization in air, not tissue, and thus do not directly quantify radiation's effect on humans. Absorbed dose is the mean energy imparted to the matter in a volume by ionizing radiation, divided by the mass of the matter in the volume. The SI unit of absorbed dose, introduced at the 15th CGPM in 1975, is the gray (Gy), which is a special name for joule per kilogram. The traditional unit is the rad, short for radiation absorbed dose, and equal to 0.01 Gy. These units are also used for air kerma.

Although absorbed dose is a useful concept, the biological effect of a given absorbed dose varies depending on the type and quality of radiation emitted by the radionuclide or external radiation field. Current ICRP terminology uses a dimensionless radiation weighting factor (w_R) to normalize for this effect, where the weighting factor ranges from 1 for photons (including x-rays and γ-rays) and electrons to 20 for α-particles. In cardiac imaging, the most common emissions are photons (nuclear cardiology), and external radiation is typically from x-rays (CT and CCA), and thus w_R is usually 1. Equivalent dose (H_T, which in most contexts has replaced the similar term *dose equivalent*) in a tissue or organ due to a radiation field is defined as the product of the absorbed dose and the radiation weighting factor. If the field is composed of types of radiation with different radiation weighting factors, then equivalent dose is determined by summing these products over the constituent radiations. Thus, equivalent dose differs from absorbed dose in that it reflects not only the energy imparted to matter by radiation but also the relative biological harm caused by the type of radiation. A special SI unit, the sievert (Sv), was adopted at the 16th CGPM in 1979 to avoid possible confusion between absorbed dose and dose equivalent and the resultant underestimation of dose equivalent.[5] The sievert is also a special name for joule per kilogram, used for doses that have been weighted to reflect the type of radiation. The traditional unit for equivalent dose is the rem, short for roentgen equivalent man, and equal to 0.01 Sv. These relationships are illustrated in Figure 1.

In addition to the absorbed dose and type of radiation, the probability of stochastic effects varies depending on the organ or tissue irradiated. A second weighting factor, the dimensionless tissue weighting factor (w_T), is used to normalize for this effect. Equivalent dose multiplied by w_T is termed *weighted equivalent dose*, properly measured in sieverts or rem. The sum of weighted equivalent dose over all organs or tissues in an individual is termed the *effective dose* (E), that is,

$$E = \sum_T w_T \times D_T = \sum_T w_T \sum_R w_R \times D_{T,R} \qquad (1)$$

where D_T, typically measured in units of mGy, represents the mean absorbed dose in tissue T from all radiations, and $D_{T,R}$ represents the mean absorbed dose in tissue T from radiation R. The older and more cumbersome term *effective dose equivalent*, supplanted by effective dose,[4] is still found in some current literature. Thus, weighted equivalent dose to a particular tissue corresponds to the contribution to E of the radiation absorbed by that tissue. Tissue-weighting factors are chosen to sum to 1 so that a uniform equivalent dose over the whole body results in an E equal to that equivalent dose, and therefore the equivalent dose to a particular organ corresponds to the E of a hypothetical scan in which each organ receives the same dose as does the particular organ.

The ICRP has offered recommended tissue weighting factors in 2 reports, their Publication 26[8] (1977) and the subsequent Publication 60[4] (1991). The highest ICRP Publication 60 w_T is that of the gonads (0.2), followed by the bone marrow, colon, lung, and stomach (each 0.12). Minor changes to ICRP Publication 60 tissue weighting factors were suggested in subsequent reports. On March 21, 2007, a comprehensive update to ICRP Publication 60 was approved; this is scheduled for publication as ICRP Publication 103.[9] Based on more current data, it introduces a new set of tissue weighting factors, summarized in Table 1. The major difference is a higher w_T for the female breast and a lower factor for the gonads. The actual dose received by a person from a given radiation exposure can be estimated by 1 of several methods, tailored to the nature of the radiation exposure.

TABLE 1. Tissue Weighting Factors in ICRP Publication 26, ICRP Publication 60, and the 2007 Draft of ICRP Publication 103

	ICRP 26[8] (1977)	ICRP 60[4] (1991)	ICRP 103[9] (2007)
Bladder	...	0.05	0.04
Bone	0.03	0.01	0.01
Brain	...	...	0.01
Breasts	0.15	0.05	0.12
Colon	...	...	0.12
Esophagus	...	0.05	0.04
Liver	...	0.05	0.04
Lower large intestine	...	0.12	...
Lungs	0.12	0.12	0.12
Ovaries/testes	0.25	0.20	0.08
Red marrow	0.12	0.12	0.12
Remainder tissues	0.30	0.05	0.12
Salivary glands	...	...	0.01
Skin	...	0.01	0.01
Stomach	...	0.12	0.12
Thyroid	0.03	0.05	0.04

Ellipses indicate no tissue weighting factor associated with organ.

Limitations in Dose Estimation

It is important to note that all reported radiation doses, both for a typical study in a population and for a particular study in a particular patient, are estimates in a statistical sense, obtained with the use of measured quantities but making numerous assumptions that may result in variation from the "true" value. For example, current radiopharmaceutical dosimetry models yield an estimate of E that is not patient-specific but rather is based on a number of assumptions, including standard patient weights and organ sizes, generic rather than patient-specific biokinetic data, and uniform radiopharmaceutical activity within organs.[10] Thus, reported doses should properly be viewed as dose estimates.[11] Although point estimates of typical doses of cardiac imaging studies have been reasonably well documented in the literature, the quantitative characterization of uncertainty in dose estimation has lagged behind and remains an important area for future investigation.

For all modalities, gender-specific dosimetry has been lacking and is only beginning to be addressed. The effect of body habitus, and obesity in particular, on dosimetry remains unclear, and dose estimation will continue to evolve as more data are available. Even so, the dose estimates here are useful in comparing different modalities and study protocols.

Nuclear Cardiology Dosimetry

Terminology and Methodology

The activity (A) of a radionuclide is the average number of spontaneous nuclear decays in a given period of time. The SI unit for activity is the becquerel (Bq), which was formally defined at the 15th CGPM as seconds^{-1} but is commonly used to mean decays per second. The traditional unit for activity is the curie, originally standardized in 1910 by Marie Curie and now equal to exactly 3.7×10^{10} Bq.

Radiation dosimetry from a study using a radiopharmaceutical is typically estimated on the basis of a mathematical biokinetic model that quantifies the distribution and metabolism of that agent in the body. Such models incorporate biokinetic data from animal and human models. They enable the determination of tissue or organ absorbed doses per unit of activity administered (D_T/A) and whole-body effective dose per unit of activity (E/A), referred to as *dose coefficients*.

Values for D_T/A are referred to here as *tissue dose coefficients*, and those for E/A are referred to as *effective dose coefficients*. A widely respected series of such models for commonly used radiopharmaceuticals has been compiled by the ICRP, drawing on the work of the Medical Internal Radiation Dose committee of the Society of Nuclear Medicine, as well as research done at Oak Ridge National Laboratories. ICRP Publication 17 (1968) and its successor ICRP Publication 53[12] (1987) contain an extensive set of

TABLE 2. Estimates of Effective Doses of Standard Myocardial Perfusion Imaging Protocols

	Injected Activity (mCi)		Effective Doses, mSv			
			From ICRP Tables		From Manufacturers' PIs	
Protocol	Rest	Stress	E_1	E_2	E_1	E_2
^{99m}Tc sestamibi rest-stress	10.0	27.5	11.3	11.4	14.6	NR
^{99m}Tc sestamibi stress only	0.0	27.5	7.9	8.0	10.0	NR
^{99m}Tc sestamibi 2-day	25.0	25.0	15.7	15.6	20.6	NR
^{99m}Tc tetrofosmin rest-stress	10.0	27.5	9.3	9.9	9.7	12.9
^{99m}Tc tetrofosmin stress only	0.0	27.5	6.6	7.1	6.7	8.8
^{99m}Tc tetrofosmin 2-day	25.0	25.0	12.8	13.5	13.7	18.3
^{201}Tl stress-redistribution	0.0	3.5	22.0	22.0	28.7 (PI 1) 9.3 (PI 2) 28.4 (PI 3)	46.6 (PI 1) NR (PI 2) 46.6 (PI 3)
^{201}Tl stress-reinjection	1.5	3.0	31.4	31.5	43.0 (PI 1) 14.0 (PI 2) 42.6 (PI 3)	69.9 (PI 1) NR (PI 2) 69.9 (PI 3)
Dual isotope ^{201}Tl-^{99m}Tc sestamibi	3.5	25.0	29.2	29.3	37.8 (PI 1) 18.4 (PI 2) 37.5 (PI 3)	NR (PI 1) NR (PI 2) NR (PI 3)
^{99m}Tc-labeled erythrocytes	22.5	0.0	5.7	5.8	2.3	NR
^{82}Rb	50.0	50.0	13.5	12.6	3.0	NR
^{13}N-ammonia	15.0	15.0	2.4	2.2	NA	NA
^{15}O-water*	29.7	29.7	2.5	2.4	NA	NA
^{18}F-FDG	10.0	0.0	7.0	7.0	NA	NA

E_1 indicates effective dose estimated from tissue dose coefficients, using ICRP Publication 60 tissue weighting factors. Calculations were performed with the use of the "splitting rule,"[4] arithmetic averaging rather than mass averaging of individual remainder organ dose contributions,[18] and upper large intestine rather than extrathoracic airways as a remainder organ, as was originally specified in ICRP Publication 60. If dose to the colon was not specified in a data source, then the average of the upper large intestine and lower large intestine doses was substituted. E_2 indicates effective dose estimated from effective dose coefficients, using ICRP Publication 60 tissue weighting factors. NR indicates not reported in PI (total body dose provided rather than effective dose); NA, not available for cyclotron-produced tracers.

*American Society of Nuclear Cardiology guidelines[16] do not prescribe a recommended dose. Stress and rest doses of 1100 MBq (29.7 mCi) used, as per European Association of Nuclear Medicine/European Society of Cardiology guidelines.[17]

dosimetry tables for a variety of radiopharmaceuticals based on these models. ICRP Publication 80[13] (1998) and the still-unpublished Addendum 5 to ICRP Publication 53[14] use more updated methodology to recalculate dosimetry for common radionuclides and correct errors in dosimetry calculations found in ICRP Publication 53. The manufacturers of radiopharmaceuticals also provide such tables in the package inserts (PIs) for these products. Some of these tables provide a total body dose rather than *E/A*. Total body dose is an older term, defined as the total radiation energy absorbed in the body divided by the mass of the body (70 kg is typically used). However, the total body dose does not account for the nonuniformity in dose distribution among body organs, and it is always lower than the effective dose.[15] Tables II and III in the online-only Data Supplement compile dose coefficients for commonly used cardiac radiopharmaceuticals from the most recent ICRP publications reporting these quantities, as well as from current manufacturers' PIs. Although for ^{99m}Tc sestamibi and tetrofosmin, separate dose coefficients are reported for injection at stress versus at rest, demonstrating modestly (8% to 22%) decreased stress doses, these data are unavailable for other agents or for injection after pharmacological stress agents.

With the use of these dose coefficients, the equivalent dose to tissue T from a radiopharmaceutical with activity A_0 can be estimated from the equation

$$H_T = (D_T/A) \times A_0 . \quad (2)$$

For each radiopharmaceutical, E can be estimated either with a set of tissue dose coefficients $\{D/A_T\}$, using

$$E_1 = \sum_T w_T \times H_T = \sum_T w_T \times (D_T/A) \times A_0 \quad (3)$$

or, alternatively, from an effective dose coefficient using

$$E_2 = (E/A) \times A_0 . \quad (4)$$

Here we use E_1 to denote an effective dose derived from tissue dose coefficients and E_2 to denote an effective dose derived from an effective dose coefficient.

Dosimetry of Nuclear Cardiology Studies

With the use of ICRP or PI dose coefficients, a set of tissue weighting factors, the radionuclide activities for a standard protocol, and Equations 3 and 4 above, one can estimate the effective dose to a typical patient of a standard cardiac radiopharmaceutical study. Table 2[16–18] summarizes E_1 and

TABLE 3. Effect of ICRP Tissue Weighting Factors w_T on Estimates of Effective Dose E_1 (mSv)

	Dose Coefficients D_T/A From ICRP Tables			Dose Coefficients D_T/A From Manufacturers' PIs		
Protocol	ICRP 26 w_T	ICRP 60 w_T	ICRP 103 w_T	ICRP 26 w_T	ICRP 60 w_T	ICRP 103 w_T
^{99m}Tc sestamibi rest-stress	7.8	11.3	10.7	8.0	14.6	12.1
^{99m}Tc sestamibi stress only	5.6	7.9	7.5	5.6	10.0	8.4
^{99m}Tc sestamibi 2-day	10.7	15.7	14.8	11.2	20.6	17.0
^{99m}Tc tetrofosmin rest-stress	5.7	9.3	8.6	7.1	9.7	8.9
^{99m}Tc tetrofosmin stress only	4.1	6.6	6.2	4.9	6.7	6.2
^{99m}Tc tetrofosmin 2-day	7.9	12.8	11.8	9.9	13.7	12.5
^{201}Tl stress-redistribution	19.2	22.0	16.9	23.5 (PI 1) 7.5 (PI 2) 23.7 (PI 3)	28.7 (PI 1) 9.3 (PI 2) 28.4 (PI 3)	21.7 (PI 1) 6.4 (PI 2) 21.4 (PI 3)
^{201}Tl reinjection	27.4	31.4	24.2	35.3 (PI 1) 11.3 (PI 2) 35.5 (PI 3)	43.0 (PI 1) 14.0 (PI 2) 42.6 (PI 3)	32.6 (PI 1) 9.5 (PI 2) 32.1 (PI 3)
Dual isotope ^{201}Tl-^{99m}Tc-sestamibi	24.2	29.2	23.7	28.6 (PI 1) 12.6 (PI 2) 28.8 (PI 3)	37.8 (PI 1) 18.4 (PI 2) 37.5 (PI 3)	29.3 (PI 1) 14.0 (PI 2) 29.0 (PI 3)
^{99m}Tc-labeled erythrocytes	4.9	5.7	5.7	2.8	2.3	1.7
^{82}Rb	10.5	13.5	12.8	3.2	3.0	2.4
^{13}N-ammonia	2.0	2.4	2.3	NA	NA	NA
^{15}O-water	1.6	2.5	2.3	NA	NA	NA
^{18}F-FDG	6.4	7.0	6.4	NA	NA	NA

NA indicates not available for cyclotron-produced tracers.

E_2 for commonly performed studies, with calculations performed with the use of ICRP Publication 60 tissue weighting factors and average radionuclide activities specified in current American Society of Nuclear Cardiology guidelines.[16] No additional radiation dose for attenuation correction is included. Performed in a minority of nuclear cardiology laboratories, attenuation correction scans performed with either radioisotope sources or low-dose CT have *E*s that are small compared with those of radionuclide studies.[19,20] Table 3 demonstrates the effect of the tissue weighting factors on effective dose, using the 3 ICRP w_T schema, and compares E_1 determined with the use of dose coefficients from ICRP tables with those from manufacturers' PIs. Organ doses for selected protocols are summarized in Table IV in the online-only Data Supplement, which lists the organs receiving the highest equivalent doses for each protocol. Figure 2 demonstrates the components (weighted equivalent doses) contributing to the total effective dose for selected protocols.

As is seen in the tables and figures, effective doses of myocardial perfusion imaging (MPI) procedures are nontrivial and vary greatly between protocols. Substantial differences exist between procedures with the use of different radiopharmaceuticals and between different procedures with the use of the same agent. While the typical effective dose of a posteroanterior chest x-ray is 0.02 mSv,[7] and the annual background radiation in the United States is 3.0 mSv,[21] typical E_1 values for MPI studies range from 2.2 to 31.5 mSv with the use of ICRP dose coefficients and ICRP Publication 60 w_T. Of the most commonly performed studies, a rest-stress ^{99m}Tc sestamibi study averages 11.3 mSv, and a rest-stress ^{99m}Tc tetrofosmin study averages 9.3 mSv. Single-injection protocols are associated with a dose that is ≈30% lower. Doses are much higher for studies using ^{201}Tl. A single-injection ^{201}Tl MPI study has an E_1 of 22 mSv. Dual isotope studies have the highest effective doses, with an E_1 of 29.2 mSv for a ^{201}Tl-^{99m}Tc sestamibi study, ≈3 times that of a single-injection protocol using a ^{99m}Tc-containing agent. The lowest doses are for positron emission tomography protocols using the cyclotron-produced radionuclides ^{13}N ammonia and ^{15}O water, for which E_1 values were 2.4 and 2.5 mSv.

Effective doses of MPI studies using the new 2007 w_T are slightly lower than those using ICRP Publication 60 w_T, as shown in Table 3 and Figure 2. The most significant factor appears to be the lower gonadal doses obtained with the new w_T, which most affects effective doses of studies incorporating ^{201}Tl.

Comparison of Doses Determined Using ICRP Versus Manufacturers' Dose Coefficients

Some notable differences exist between effective doses estimated with the use of ICRP dose coefficients and those estimated with the use of dose coefficients provided in PIs, as illustrated in Table 3 and Figure 3. Most PIs were initially issued at the time of approval of a radiopharmaceutical, and dosimetry information included in subsequent revisions has not been updated to reflect new biokinetic data or changes in the ICRP dosimetry system. Determination of E_2 is not possible from the PIs for ^{99m}Tc sestamibi, ^{99m}Tc-labeled erythrocytes, ^{82}Rb, and 1 of the 3 manufacturers of ^{201}Tl. Each of these reports total body dose per unit activity rather than effective dose per unit activity. Future revisions of these PIs should incorporate effective dose coefficients.

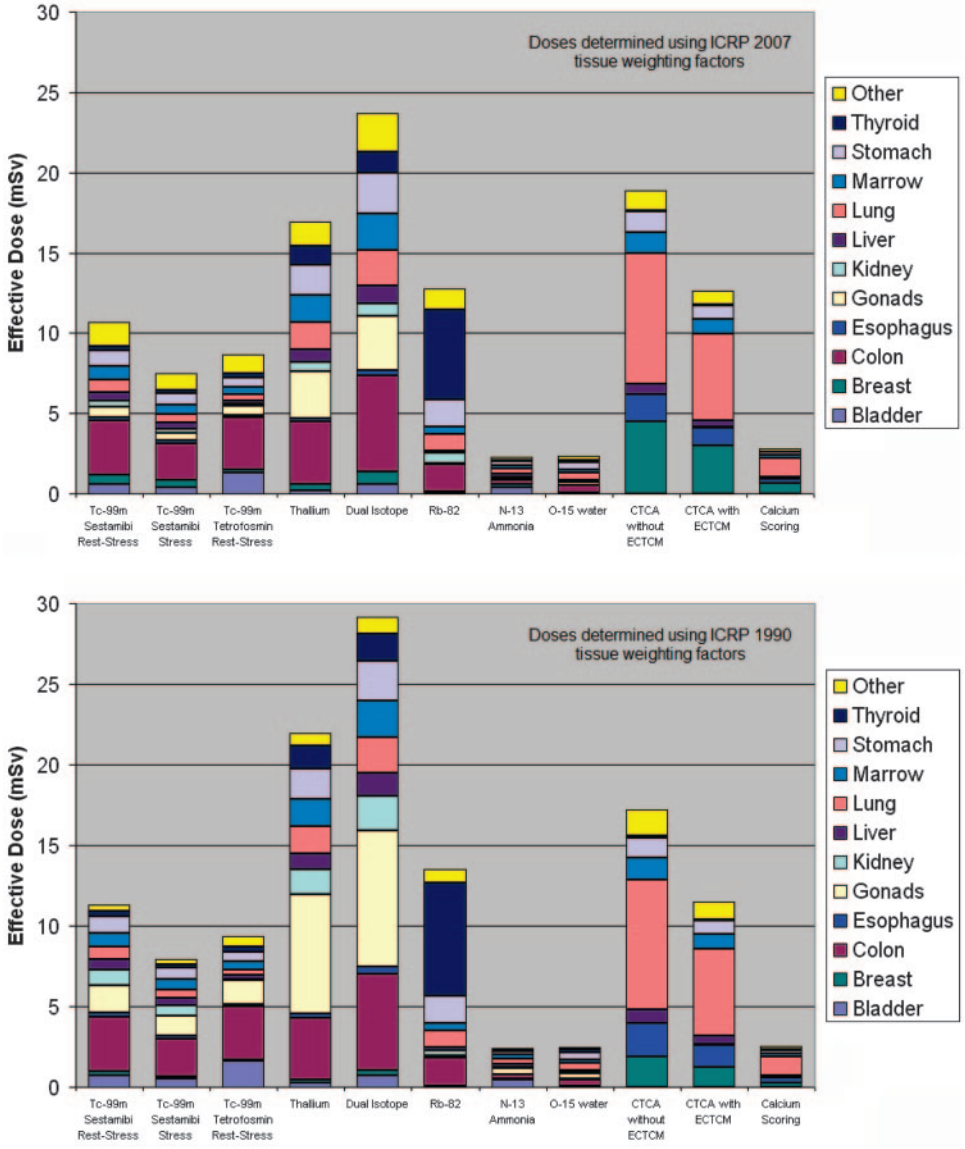

Figure 2. Estimated effective doses and weighted organ equivalent doses from some standard cardiac radionuclide studies. Top, Doses determined using ICRP Publication 103 (2007) tissue weighting factors. Bottom, Doses determined using ICRP Publication 60 (1990) tissue weighting factors. CTCA indicates 64-slice computed tomography coronary angiogram; CaSc, calcium scoring; and ECTCM, ECG-controlled tube current modulation. Calculations were performed with ImpactDose (VAMP GmbH, Erlangen, Germany); for Siemens Sensation 64 scanner with retrospective gating, voltage was 120 kVp, pitch 0.2, and scan length 15 cm. For CTCA, slice thickness was 0.6 mm and tube current-time product was 165 mAs; ECTCM was simulated by reducing tube current by a third, to 110 mAs. For CaSc, collimation was 20×1.2 mm, and tube current-time product was 27 mAs. Effective doses (E_1) correspond to third and second numeric columns in Table 3, respectively. Doses shown are arithmetic means of doses to standardized male and female phantoms.

For ^{99m}Tc sestamibi rest-stress imaging, good agreement exists between E_1 from ICRP (11.3 mSv) and PI (14.6 mSv for 4.8-hour urinary void, 13.5 mSv for 2-hour void) dose coefficients. The higher E_1 with the longer void time is primarily due to the higher equivalent dose to the bladder wall (41 versus 21 mSv) and demonstrates the potential dose-reduction benefit of hydration and early micturition after radiopharmaceutical administration. Much of the difference between ICRP- and PI-derived E_1 is due to an idiosyncrasy in the methodology of ICRP Publication 60 for determining dose to "remainder" organs, which was later amended.[22] For ^{99m}Tc tetrofosmin, even closer agreement exists between ICRP- and PI-derived E_1 values, which are 9.3 and 9.7 mSv, respectively.

^{201}Tl dosimetry varies markedly between manufacturers' PIs. E_1 for a 3.5-mCi injection determined with dose coefficients from ICRP Publication 53 Addendum 5 and PIs 1, 2, and 3 are 22.0, 28.7, 9.3, and 28.4 mSv, respectively. When we examine the dose coefficients in PI 2, included in Table III in the online-only Data Supplement, no doses are listed for many organs (this is noted as well for ^{99m}Tc-labeled erythrocytes), and dose coefficients are much lower in general than for the other sources of data. In contrast, *E/A* for ^{201}Tl is much higher in PIs 1 and 3 (0.36 mSv/MBq) than in ICRP

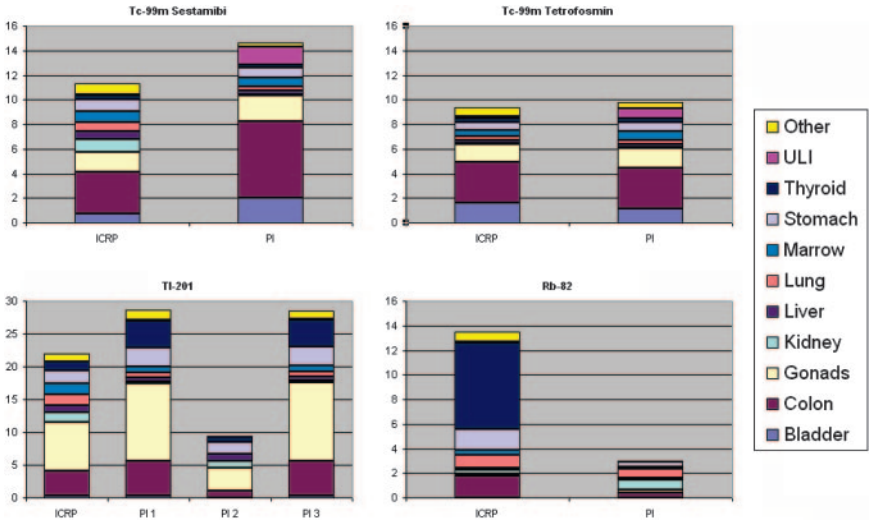

Figure 3. Comparison of estimated effective doses (mSv) for standard myocardial perfusion imaging protocols, determined with the use of ICRP and manufacturers' PI dose coefficients. Weighted equivalent doses were determined with the use of ICRP Publication 60 tissue weighting factors. ULI indicates upper large intestine.

Publication 53 Addendum 5 (0.17 mSv/MBq), resulting in a discordance between PI-derived E_1 and E_2 and extremely high E_2 for standard protocols, ie, 47 mSv. Data sources cited in these PIs date back to the 1980s, and even with the use of ICRP Publication 26 tissue weighting factors, the discordance between E_1 and E_2 remains. In sum, 2 PIs suggest a ^{201}Tl effective dose even greater than that from ICRP data, and a third PI (reporting limited organ data and no effective dose coefficient) suggests a much lower effective dose. It appears that ^{201}Tl dosimetry requires revisiting, and PIs should be updated, which will result in lower effective dose coefficients for 2 manufacturers if ICRP Publication 53 Addendum 5 dosimetry is confirmed.

Another radiopharmaceutical for which significant differences exist between ICRP and PI dosimetry is ^{82}Rb. For ^{82}Rb (in contrast to ^{201}Tl, for which both D_T/A and E/A were recalculated in the recent ICRP Publication 53 Addendum 5), the ICRP has not published updated D_T/A since 1987,[12] although E_1 values derived from these older D_T/A are consistent with E_2 values derived with the use of the E/A from ICRP Publication 80 (2000).[13] PI dose coefficients for ^{82}Rb were determined with the use of a limited number of human subjects. The most marked difference in organ doses between ICRP and PI data is the dose to the thyroid. The weighted equivalent dose to the thyroid for a standard 100 mCi protocol is 7 mSv with the use of ICRP Publication 53 data and 0 mSv with the use of PI data, which does not include a thyroid dose coefficient. Accurate ^{82}Rb dosimetry is essential, particularly because ^{82}Rb generator installations are increasing, and thus reevaluation of dose coefficients is needed.

Strategies to Minimize Dose in Cardiac Nuclear Imaging

Dosimetric considerations have important implications for the selection of MPI protocols. In 2002, 35% of the 9.3 million MPI studies performed in the United States used ^{201}Tl, with 86% of these being dual isotope studies. The use of these high-dose protocols appears to be increasing, with 30% of studies in 2002 being dual isotope compared with 19% in 1997.[1] Dual isotope studies are particularly common in the outpatient setting, in which they are used in 36% of all MPI studies, perhaps because of the relatively fast patient throughput. However, the radiation dose of studies employing ^{201}Tl, especially dual isotope MPI, is among the highest of all medical diagnostic tests. Thus, ALARA considerations appear to favor the use of ^{99m}Tc agents rather than ^{201}Tl. Nevertheless, in some cases a protocol employing ^{201}Tl is preferred, for example, in cases in which a viability assessment is desirable or in patients with a history of ^{99m}Tc images obfuscated by increased gastrointestinal tracer uptake. Although dose to the patient is minimized with the use of ^{99m}Tc agents, the high activities used in these protocols has resulted in nuclear cardiology technologists receiving some of the highest radiation exposures among nuclear medicine personnel.[23]

A number of strategies can be used to minimize dose in cardiac nuclear imaging (Table 4). One appealing but not widely utilized approach to lower radiation dose is the use in patients with low pretest probability of disease of stress-first/stress-only protocols employing ^{99m}Tc sestamibi or tetrofosmin, often in conjunction with attenuation correction. Only 9% of sites performing nuclear cardiology procedures in the United States in 2002 offered single-injection protocols, and only 4% of studies actually used only a single injection of a ^{99m}Tc agent.[1] Radiation dose from stress-first imaging is even lower than in 2-injection ^{99m}Tc studies, study time is low for patients requiring a single injection, and more patients can be imaged per gamma camera per day. However, diagnostic performance and prognostic value have not been evaluated as extensively for stress-only imaging as for protocols incorporating stress and rest imaging. Moreover, stress-first/stress-only protocols may necessitate a second visit to the nuclear laboratory, and its attendant second radiation dose, for some patients. This can be minimized by communication of clinical information between the referring physician and the nuclear laboratory in sufficient detail to enable accurate pretest risk stratification and selection of patients for stress-first/stress-only protocols.

Cardiac CT Dosimetry

CT scanners used for cardiac applications are based on 2 types of architecture. Whereas earlier studies were performed

TABLE 4. Strategies to Minimize Radiation Dose to Patients From Cardiac Diagnostic Imaging

SPECT/PET
^{99m}Tc agents preferred when possible in SPECT
Consider stress-first/stress-only protocol for patients with low pretest probability of stress perfusion defect
Minimize activity (mCi) to that needed to obtain good image quality with high degree of confidence
Consider lower activity (mCi) in smaller patients
For CT attenuation correction, minimize tube current
Hydrate after imaging and encourage early micturition
CT
Employ ECG-controlled tube current modulation when possible (low heart rate, regular rhythm)
Use β-blockers to lower heart rate, which improves efficacy of ECG-controlled tube current modulation
Minimize scan length
Prospectively gate calcium scoring scan
Match tube current to patient habitus for calcium scoring and CTCA
Consider avoidance of CTCA if calcium scoring scan reveals widespread, heavy coronary calcification
CCA
Employ slowest fluoroscopy and fluorography frame rates that maintain diagnostic image quality
Minimize fluoroscopy and fluorography time
Use least amount of image magnification needed for accurate interpretation
Minimize distance from patient to image detector and x-ray tube
Optimize beam collimation
Minimize number of views
Shield sensitive organs (eg, gonads)
Use highest acceptable kV to maintain lowest possible mA
Omit left ventriculography if the diagnostic information is available from other tests

SPECT indicates single photon emission computed tomography; PET, positron emission tomography.

with electron beam CT scanners, this technology has been largely supplanted in recent years by multidetector-row CT scanners, which have higher spatial resolution, and our discussion here is limited to the latter. As multidetector-row CT has progressed, a particular set of terminology has developed for its radiation dosimetry, reviewed in greater detail in a number of sources.[24–27]

Terminology and Methodology

The dose profile [D(z)] for a CT scanner is a mathematical description of the dose as a function of position on the z axis (perpendicular to the tomographic plane). The CT dose index (CTDI), measured in units of grays, is the area under the radiation dose profile for a single rotation and fixed table position along the axial direction of the scanner, divided by the total nominal scan width or beam collimation, that is,

$$CTDI = \frac{\int_{-\infty}^{\infty} D(z)dz}{N \times T}, \quad (5)$$

where N is the number of tomographic sections produced simultaneously in the rotation of the x-ray tube, and T the section thickness. CTDI is difficult to measure, and therefore in practice $CTDI_{100}$ is generally determined, which represents the integrated radiation dose from acquiring a single scan over a length of 100 mm. Air kerma or exposure is measured with the use of a pencil ionization chamber placed in a cylindrical polymethylmethacrylate phantom (Figure 4), at both the phantom's center and its periphery, and converted to a dose. By convention, a phantom 16 cm in diameter is used to model the head, and a 32-cm phantom is used to model the body. Weighted CTDI ($CTDI_w$) estimates, from $CTDI_{100}$ measurements, the average radiation dose to a cross section of a patient's body. It is determined with the equation

$$CTDI_w = \tfrac{2}{3}\, CTDI_{100} \text{ at periphery} + \tfrac{1}{3}\, CTDI_{100} \text{ at center}. \quad (6)$$

An important CT-specific dosimetry term is the volume CTDI ($CTDI_{vol}$). This quantity, established by the International Electrotechnical Commission in 2002,[29] represents the average radiation dose over the volume scanned. It is determined for helical scans from the $CTDI_w$ by the equation

$$CTDI_{vol} = CTDI_w/\text{pitch} = CTDI_w \times \frac{\text{total nominal scan width}}{\text{distance between scans}} \quad (7)$$

$CTDI_{vol}$ can be used in turn to determine the dose-length product (DLP). Measured in units of mGy · cm, DLP reflects the integrated radiation dose for a complete CT examination and is calculated by

$$DLP = CTDI_{vol} \times \text{length irradiated}. \quad (8)$$

DLP can be related to E by the formula

$$E = E_{DLP} \times DLP, \quad (9)$$

where E_{DLP}, measured in units of mSv/(mGy · cm), is a body region–specific conversion factor. The most commonly used E_{DLP} values are those of the European Guidelines on Quality Criteria for Computed Tomography,[26] although newer values are reported in the 2004 CT Quality Criteria (Table 5).[27] These E_{DLP} values are determined by Monte Carlo methods, averaged for multiple scanners. E_{DLP} values based on ICRP 2007 tissue weighting factors are not yet available.

Although the aforementioned system of nomenclature for CT is the present standard, the International Atomic Energy Agency has recently adopted a different system of nomenclature based on the precise terminology of the ICRU, which may eventually replace some currently standard terms.[30,31] The basic idea reflected is that in practice, dosimetry equipment properly measures an air kerma rather than an absorbed dose. An example of this change in nomenclature is the substitution of the term *CT dose index* by *CT air kerma index* and the term *dose-length product* by *air kerma-length product*.

Estimating Dose in Practice

The actual dose received by a patient from a given CT examination is commonly estimated by 1 of 3 approaches:

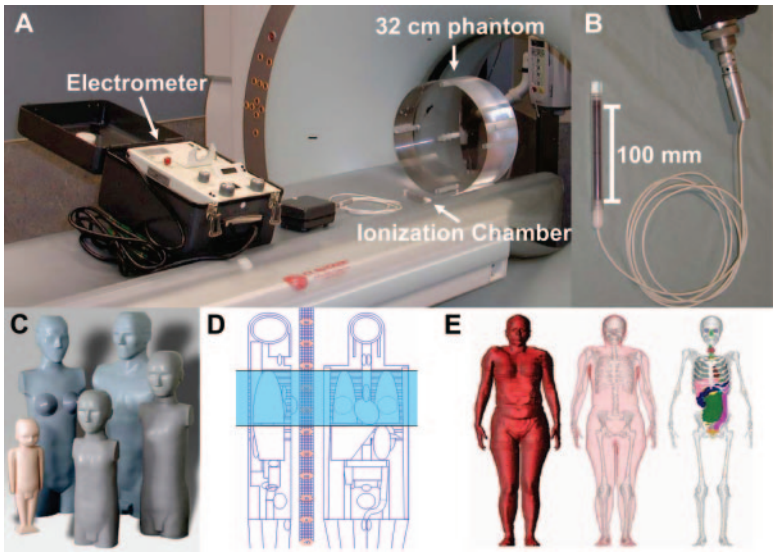

Figure 4. CT dosimetry measurement tools. A, Electrometer and ionization chamber set up to take CTDI measurements in a 32-cm polymethylmethacrylate "body" phantom. B, Close-up of the 100-mm pencil ionization chamber. C, Physical anthropomorphic phantoms (ATOM, CIRS Inc, Norfolk, Va). Reproduced with permission from CIRS Inc. D, Geometric mathematical phantom used in Monte Carlo simulations. E, Voxel mathematical phantom used in Monte Carlo simulations. Reproduced from Dimbylow,[28] with permission from the publisher. Copyright © 2005, IOP Publishing.

(1) with calculations based on physical measurements made in physical phantoms (Figure 4); (2) with $CTDI_{vol}$ or DLP values provided on the scanner console in conjunction with Equations 8 and 9; or (3) with Monte Carlo simulations. Physical measurements may be made with the use of ionization chambers, lithium fluoride or calcium fluoride thermoluminescent dosimeters, metal oxide semiconductor field effect transistors, film, aluminum oxide crystals, or other solid state detectors. In fact, the $CTDI_{vol}$ and DLP reported on a scanner console are typically determined from measurements made with a pencil ionization chamber in the specific scanner model. Monte Carlo simulations assume a mathematical patient phantom and model photon transport through this simulated patient. The 2 most widely used models are those developed by the Gesellschaft für Strahlen- und Umweltforschung and the United Kingdom's National Radiological Protection Board, now part of the Health Protection Agency. Software is available with data derived from each of these models to estimate patient doses for current scanners with the use of particular scan protocols (Table 6). With these Monte Carlo method–based programs, parameters such as the scanner, tube current, tube voltage, pitch, and scan area in the simulation can be matched to those in an actual examination, enabling realistic simulation of radiation dosimetry in a clinical CT examination. Current software uses geometrical phantoms, modeling organs as simple geometrical shapes. Newer, more anatomically detailed voxel phantoms have been developed and offer the potential of more accurate dosimetry, tailored to body habitus. Regardless of the phantom used for dosimetry, be it geometric, voxel, or a physical phantom, one must be aware that the effective dose accurately estimates the radiation dose to the patient only insofar as the phantom is reflective of the patient's anatomy. Although it appears that for a given scanner and set of scan parameters, heavier patients will have lower *E*,[32] higher tube currents are typically employed for obese patients in CTCA. Unfortunately, current literature and software inadequately address the relationship between habitus and effective dose for imaging studies employing ionizing radiation; further research in this area is essential.

Dose From Cardiac CT

Several studies have estimated *E* of cardiac CT (Table 7).[33–53] Mean *E* for calcium scoring using retrospective gating ranges from 1.0 to 6.2 mSv, depending on the protocol and scanner used. Mean *E* is lower using prospective gating, with a range from 0.5 to 1.8 mSv, although this does not include any 64-slice studies. Mean *E* for CTCA in the studies in Table 7 ranges from 4.0 to 21.4 mSv. Studies involving more recent, eg, 64-slice, scanners typically report higher *E*s. Such scanners, with higher spatial resolution, use increased tube currents that translate to higher doses. As illustrated in Figure 2, whereas CTCA and rest-stress ^{99m}Tc sestamibi MPI have similar *E*, the organ contributions to this stochastic risk

TABLE 5. Relation Between DLP and E: $E=E_{DLP}\times DLP$

	E_{DLP}, mSv·mGy^{-1}·cm^{-1}		
Region of Body	European Guidelines 2000[26]	2004 CT Quality Criteria[27] Appendix A	2004 CT Quality Criteria[27] Appendix C
Head	0.0023	0.0023	0.0021
Neck	0.0054	0.0054	0.0059
Chest	0.017	0.019	0.014
Abdomen	0.015	0.017	0.015
Pelvis	0.019	0.017	0.015

TABLE 6. Some Software Programs to Estimate Effective Dose From Radiological Studies

Modality	Program	Web Site
SPECT/PET	OLINDA/EXM*	http://www.doseinfo-radar.com/OLINDA.html
SPECT/PET	CDI3	http://www.fda.gov/cdrh/ohip/organdose.html
CT (GSF data)	ImpactDose†	http://www.vamp-gmbh.de/software/impactdose.php
CT (GSF data)	CT-Expo	http://www.mh-hannover.de/1604.html
CT (NRPB data)	CTDosimetry.xls	http://www.impactscan.org/ctdosimetry.htm
CT	CTDOSE	http://www.mta.au.dk/ctdose/index.htm
Fluoroscopy	WinODS	http://www.rti.se/download_software/winods_demo_instr.htm
Fluoroscopy	PCXMC	http://www.stuk.fi/sateilyn_kayttajille/ohjelmat/PCXMC/en_GB/pcxmc/
Fluoroscopy	XDOSE	http://www.hpa.org.uk/radiation/publications/software/sr262.htm
General purpose	MCNP/MCNPX	http://mcnpx.lanl.gov/
General purpose	Geant4	http://geant4.web.cern.ch/geant4/

SPECT indicates single photon emission computed tomography; PET, positron emission tomography; GSF, Gesellschaft für Strahlen- und Umweltforschung; and NRPB, National Radiological Protection Board.

*Formerly MIRDOSE.

†Formerly WinDose.

(weighted equivalent doses) are markedly different. The organs receiving the highest equivalent doses in CTCA are the female breasts, lungs, liver, and esophagus. One 16-slice study reported a breast equivalent dose from CTCA of 55.6 mSv, which was reduced to 27.1 mSv with ECG-controlled tube current modulation (ECTCM),[43] underscoring the importance of using ECTCM whenever appropriate. In this method, tube current is reduced to a small fraction of its maximum value during portions of the cardiac cycle, eg, early systole, in which image data are not typically used for interpretation, because of the presence of coronary motion. ECTCM has been observed to result in dose reductions near 50% in the best cases.

Strategies to Minimize Dose From Cardiac CT

A number of techniques can be used to minimize dose from cardiac CT. For calcium scoring, prospective gating is recommended. Ideally, the noncontrast calcium scoring scan should be examined before proceeding with CT angiography because widespread calcification may render many coronary segments difficult to interpret. For angiography, ECTCM should be employed when it is expected that multiple reconstructions at different portions of the cardiac cycle will not be necessary to interpret the images. This is generally the case for patients with regular rhythm, little or no ectopy, and well-controlled heart rate after administration of an AV nodal blocking agent such as metoprolol. β-Blockers play an important role in dose reduction in addition to their role in improving image quality by decreasing coronary artery velocity. The lower the heart rate, the greater is the reduction in effective dose from ECTCM, as was demonstrated by Jakobs et al.[54] Thus, all patients for whom ECTCM is employed should be rate controlled to $\leq$55 bpm, if feasible. Another component of dose reduction is minimization of scan length with the use of the scout and, when available, calcium scoring scan.[55] Yet another important consideration is the optimization of tube current and voltage. *E* increases linearly with tube current,[39] and therefore tube current should be minimized to the lowest level expected to yield good image quality for the particular scanner and patient habitus. One approach considered by Hausleiter et al[46] is reducing tube voltage from the standard 120 kV to 100 kV because dose varies approximately with the voltage squared. In a subgroup analysis, for a 64-slice scanner, 50 patients studied with tube voltage of 120 kV were compared with 30 patients with tube voltage of 100 kV. In all patients, ECTCM was employed. Although the mean *E* was 43% less, the percentage of unevaluable coronary artery segments was lower in the 100-kV group, which was attributed to greater vascular opacification from an increase in the photoelectric effect and a decrease in Compton scattering. This approach requires further validation before its adoption in practice because the numerous studies evaluating the diagnostic performance of 64-slice CT angiography uniformly use a tube voltage of 120 kV.

Current frontiers in manufacturers' development of lower-dose cardiac CT scanners revisit 2 features found in previous generations of scanners: multiple x-ray sources and prospective gating. Multiple sources enable increasing the pitch of a scan, ie, less overlap between gantry rotations, and correspondingly a lower dose. Interestingly, this may result in better dose reduction at higher heart rates and obviate the need for β-blockade in many patients; 1 recent study showed that increasing pitch with a dual-source CT reduced $CTDI_{vol}$ by 25% at a heart rate of 60 bpm (0.265 pitch), 44% at 78 bpm (0.36 pitch), and 57% at 100 bpm (0.46 pitch) compared with a standard protocol with a pitch of 0.2.[56] Another possibility for lowering the dose in cardiac CT is the employment of prospective gating to only acquire images during diastasis, combined with "step-and-shoot" nonspiral scanning[57] or longer detector arrays (eg, 256 detectors) enabling nonspiral whole organ imaging. The sensitivity, specificity, and dosimetry of such strategies remain to be established, but this technology is advancing rapidly, and multiple CT scanner manufacturers have recently announced the release of step-and-shoot algorithms.

TABLE 7. Studies Reporting Effective Dose in CTCA

				Mean Effective Dose Estimates, mSv					
				CTCA			Calcium Scoring		
Study	Slices	Vendor	Method	Without ECTCM	Mixed	With ECTCM	Without ECTCM	With ECTCM	Prospective Gating
Hunold et al[33]	4	Siemens	TLD-ARP (low)	6.7♂ 8.1♀	...	...	3.0♂ 3.6♀	...	1.5♂ 1.8♀
	4	Siemens	TLD-ARP (high)	10.9♂ 13.0♀	...	...	5.2♂ 6.2♀	...	...
McCollough[25]	4	... (1st)	Multiple	9.0	...	...	2.5	...	0.9
	4	... (2nd)	Multiple	12.0	...	...	4.5	...	1.1
Poll et al[34]	4	Siemens	DLP	8.3♂ 11.0♀	...	4.0♂ 5.4♀	1.9♂ 2.5♀	1.2♂ 1.6♀	...
	4	Siemens	TLD-ARP	10.3♂ 12.7♀	...	4.6♂ 5.6♀	2.4♂ 2.9♀	1.5♂ 1.8♀	...
Hacker et al[35]	12	Siemens	...	...	...	4.3	...	...	...
Coles et al[36]	12	Siemens	CTDosimetry.xls	14.2	...	...	4.1	2.6	...
	16	Siemens	CTDosimetry.xls	15.3	...	...	...	...	...
Flohr et al[37]	16	Siemens	WinDose	7.1♂ 10.5♀	...	4.3♂ 6.4♀	2.2♂ 3.1♀	...	0.45♂ 0.65♀
Garcia et al[38]	16	Philips	DLP	...	8	...	...	...	...
Gerber et al[39]	16	Siemens	Modified DLP	11.3	...	8.1	...	...	...
Hoffmann et al[40]	16	Philips	...	4.9	...	8.1	...	...	...
Nawfel and Yoshizumi[41]	16	GE	MOSFET-CIRS	20.6	...	...	...	...	...
	16	Siemens	MOSFET-CIRS	18.8	...	...	...	...	...
Sato et al[42]	4	Siemens	...	4-5	...	...	...	...	...
	16	Toshiba	...	7-8	...	...	...	...	...
Trabold et al[43]	16	Siemens	TLD-ARP	8.1♂ 10.9♀	...	4.3♂ 5.6♀	2.9♂ 3.6♀	1.6♂ 2.0♀	...
Gaspar et al[44]	40	Philips	Modified DLP	9.9	...	...	...	...	...
Caussin et al[45]	64	Siemens	...	...	8.4	...	...	...	...
Hausleiter et al[46]	16	Siemens	DLP	10.6	...	6.4	...	...	...
	64	Siemens	DLP	14.8	...	9.4	...	...	...
Ghostine et al[47]	64	Siemens	DLP	...	...	7	...	...	...
Leber et al[48]	64	Siemens	...	...	...	10-14	...	...	...
Mollet et al[49]	64	Siemens	WinDose	15.2♂ 21.4♀	...	...	...	1.3♂ 1.7♀	...
Muhlenbruch et al[50]	64	Siemens	...	13.6♂ 17.2♀	...	...	...	...	...
Nikolaou et al[51]	64	Siemens	WinDose	8-10	...	...	...	...	...
Pugliese et al[52]	64	Siemens	...	15♂ 20♀	...	...	...	...	...
Raff et al[53]	64	Siemens	...	13♂ 18♀	...	...	...	...	...

Ellipses (. . .) indicate not specified; TLD-ARP, thermoluminescent dosimeters in an Alderson-Rando phantom; DLP, derived from dose-length product on scanner console; ♂, male; ♀, female; and MOSFET-CIRS, metal oxide semiconductor field effect transistors in a CIRS Phantom.

Coronary Angiography Dosimetry

Terminology and Methodology

Terminology used to quantify dose in CCA addresses dose both to the whole body and to the skin (Figure 5).[58] Neither is reflected adequately by fluoroscopy time, which does not incorporate information about dose rate, entrance ports, or fluorography (ie, digital or cine film acquisition).[59] Contemporary fluoroscopic equipment used in CCA measures air kerma with the use of an ionization chamber incorporated into the x-ray equipment and reports the dose-area product (DAP), equal to air kerma (dose) multiplied by x-ray beam cross-sectional area. DAP is independent of the distance from the x-ray source because the decrease in dose with distance parallels the increase in area.[60]

As in CT, methods to determine E from CCA can be divided into 3 general approaches: (1) measurements in physical anthropomorphic phantoms, (2) multiplying DAP by a conversion factor, and (3) Monte Carlo simulation programs (Table 6). Multiple sources of conversion factors exist, the most widely used being those of the National Radiological Protection Board.[61] These factors vary depending on the radiographic view. Because coronary angiographic sequences are more or less standardized for most cardiac catheterization laboratories, a single average conversion factor is sometimes used for convenience to calculate E from DAP. For example, Lobotessi et al[62] used National Radiological Protection Board tables and DAP to calculate E in a cohort of patients. Mean DAP was 58.3 Gy · cm^2, and mean E was 12.9 mSv, giving an average conversion factor of 0.22 mSv/(Gy · cm^2). The Swedish Radiation Protection Authority[63] has published a conversion factor of 0.18 mSv/(Gy · cm^2), and the range reported by others varies from 0.12 to 0.26 mSv/(Gy · cm^2),[64] indicating the difficulty with using this approach to estimate E reliably.

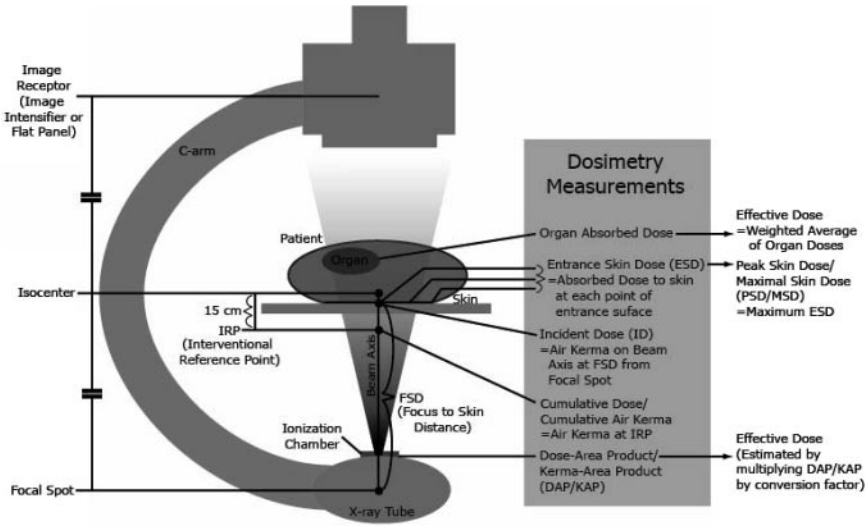

Figure 5. Fluoroscopy/fluorography dosimetry terminology. Adapted from "Avoidance of Radiation Injuries From Medical Interventional Procedures,"[58] with permission from the ICRP. Copyright © 2000, International Commission on Radiological Protection.

Quantities used to measure the risk of skin injury include peak skin dose and cumulative dose.[59] Peak skin dose, also termed *maximum skin dose* and measured in grays, is the highest absorbed dose received by any location on the patient's skin, including both incident and back-scattered radiation. Although thought to be the best predictor of skin injury, peak skin dose is difficult to measure in practice. Cumulative dose or cumulative air kerma is the total air kerma during a procedure, typically measured at the interventional reference point, the point on the x-ray beam axis lying 15 cm from the imaging system's isocenter on the x-ray tube side.[65]

Dose From Conventional Coronary Angiography

Reports of the mean E in conventional diagnostic CCA vary widely in the literature, from 2.3 to 22.7 mSv (Table 8).[66–76] The United Nations Scientific Committee on the Effects of Atomic Radiation cites a typical value of ≈7 mSv.[77] Coronary, peripheral, and electrophysiological interventional procedures with long fluoroscopy times may deliver radiation

TABLE 8. Studies Reporting Effective Dose of Conventional Coronary Angiography in Nonpediatric Populations

			Mean Effective Dose Estimate, mSv					
Study	Year	Group	CA	CA±PTCA	PTCA	CA+PTCA	ICS	CA+ICS
Karppinen et al[66]	1995	. . .	. . .	10.6	. . .	. . .	. . .	. . .
Leung and Martin[67]	1996	. . .	3.1	. . .	. . .	. . .	. . .	. . .
Broadhead et al[68]	1997	Room A	9.4	. . .	14.2	. . .	. . .	. . .
		Room B	4.6	. . .	10.2	. . .	. . .	. . .
Betsou et al[69]	1998	. . .	5.6	. . .	6.9	9.3	9	13
Harrison et al[70]	1998	. . .	3.4	. . .	. . .	. . .	. . .	. . .
Neofotistou et al[71]	1998	. . .	4.6-15.8	. . .	. . .	5.4–41.0	. . .	. . .
Katritsis et al[72]	2000	. . .	5	. . .	6.6	13.6	10.2	16.7
Lobotessi et al[62]	2001	. . .	13.2	. . .	. . .	. . .	. . .	. . .
Delichas et al[73]	2003	Hospital A	22.7	. . .	30.5	. . .	. . .	. . .
		Hospital B	17.9	. . .	14.7	. . .	. . .	. . .
Efstathopoulos et al[74]	2003	. . .	5	. . .	...	14.8	. . .	. . .
Hunold et al[33]	2003	. . .	2.3	. . .	. . .	. . .	. . .	. . .
Sandborg et al[64]	2004	Femoral	6.8	. . .	. . .	8.6	. . .	. . .
		Radial	9.2	. . .	. . .	13.5	. . .	. . .
Stisova[75]	2004	Workplace A1	8.8	. . .	. . .	. . .	. . .	. . .
		Workplace A2	3.6	. . .	. . .	9.7	. . .	. . .
		Workplace B	7.9	. . .	. . .	15.3	. . .	. . .
		Workplace C	2.7	. . .	. . .	5.7	. . .	. . .
Coles et al[36]	2006	. . .	5.6	. . .	. . .	. . .	. . .	. . .
Vijayalakshmi et al[76]	2007	. . .	4.4	. . .	. . .	. . .	. . .	. . .

CA indicates coronary angiography; PTCA, percutaneous transluminal coronary angioplasty; ICS, intracoronary stenting; and ellipses (. . .), not specified.

doses 3 to 5 times this level. Dose in catheter angiography is highly dependent on operator experience,[78] workload,[79] use of radiation-reducing techniques,[80] procedural complexity,[81] and catheterization laboratory equipment.[75]

Coronary angiography and interventions from radial artery access have been shown to be longer and associated with increased dose compared with procedures from femoral access routes.[64] Typically, fluorography ("cine") contributes most of the radiation dosage and the fluoroscopic portion of the procedure less than half for diagnostic cardiac catheterizations. Leung and Martin[67] found that the fluoroscopy DAP contribution varied from 28% to 40% among 6 cardiologists but that procedures involving right heart catheterization and coronary bypass grafts were associated with substantially higher DAP with ≥50% of the dose coming from fluoroscopy. In this study, the average *E* from fluoroscopy in patients undergoing left heart catheterization was 1.1 mSv, of an average total *E* of 3.1 mSv. Fluoroscopy and angiography in the left anterior oblique view are associated with significantly greater radiation dose than those in the posterior-anterior or right anterior oblique views.[67] The left anterior oblique views tend to be more steeply angulated, with a subsequently more oblique and longer course of x-rays through the thorax than in right anterior oblique views. These angled views are also associated with a greater source-to-target distance. Automatic brightness controls in cardiac catheterization laboratory systems increase the intensity of x-rays when attenuation and source-to-target distances increase, thus increasing patient dose.

One issue relevant to catheterization procedures not seen with CT and radionuclide imaging is the potential for radiation-induced skin injury. Coronary interventions and certain electrophysiological procedures are sometimes complex with long fluoroscopy times using few views, and there have been numerous reported cases of skin injury. Dose rates of catheterization laboratory x-ray units are relatively high, and various skin injuries ranging from transient erythema to necrosis and malignancy may occur deterministically,[82] each with a typical threshold skin dose and time course (Table 9). Dose should be monitored carefully for complex and repeat procedures. According to Hansson and Karambatsakidou,[83] the maximum permissible DAP for preventing skin injury is 530 Gy · cm^2 during CCA and 250 Gy · cm^2 during percutaneous coronary intervention to the left anterior descending coronary artery.

Strategies to Minimize Dose From Fluoroscopy

The wide variation in reported *E* from diagnostic CCA underscores the importance of optimizing technique to minimize dose. A variety of techniques should be used toward this goal. These have been reviewed in detail elsewhere[60] and are summarized in Table 4. The risk of skin injuries can be minimized by varying the radiographic projection during a procedure and by the use of real-time skin dose monitoring.

Dose and Biological Risks

The significance of measuring radiation doses comes from the relationships between dose and risks of deterministic and stochastic effects. In cardiac imaging, the only deterministic effect that occurs with any frequency in patients is skin injury, discussed above. Stochastic risks of potential concern include heritable genetic effects and cancer. Risks for all classes of genetic diseases occur at a rate estimated at 0.30% to 0.47% per Gy per first-generation progeny.[21] Even with the highest gonadal doses found in cardiac imaging, in ^{201}Tl scintigraphy with a testicular absorbed dose of ≈60 mGy (Table IV in the online-only Data Supplement, but compare Thomas et al[84]), this would correspond to a risk of genetic diseases of only 0.02% to 0.03% per first-generation progeny.

Physicians' major radiation-related concern relating to cardiac imaging is iatrogenic malignancy. Ionizing radiation causes numerous types of DNA damage, and it is hypothesized that multiply damaged sites, such as double-strand breaks, are oncogenic.[85] For the type of radiation used in cardiac imaging, ie, low levels (≤100 mSv) of low linear energy transfer ionizing radiation, the relationship between dose and lifetime attributable risk of cancer is a controversial one. Many but not all organizations offering expert opinions maintain that the linear no-threshold model, whereby the risk of cancer proceeds in linear fashion with no lower threshold, provides the most reasonable description of this relationship.[86] A National Academies committee affirming this position has developed risk models to estimate radiation-attributable cancer risk as a function of age and gender. As illustrated in Figure 6, risk falls off with age and is typically

TABLE 9. Skin Injuries Occurring After Fluoroscopy

Effect	Threshold,* Sv	Approximate Onset
Early transient erythema	2	Hours
Main erythema	6	10 d
Late erythema	15	6–10 wk
Temporary epilation	3	3 wk
Permanent epilation	7	3 wk
Dry desquamation	14	4 wk
Moist desquamation	18	4 wk
Secondary ulceration	24	>6 wk
Ischemic dermal necrosis	18	>10 wk
Dermal atrophy (first phase)	10	>14 wk
Dermal atrophy (second phase)	10	>1 y
Induration (invasive fibrosis)	10	Not available
Telangiectasia	10	>1 y
Late dermal necrosis	>12?	>1 y
Skin cancer	Unknown	>5 y

Adapted from Hirshfeld et al[60] with permission from the publisher. Copyright © 2005, The American Heart Association.

* May vary depending on patient-specific characteristics.

	Age at Exposure (years)										
	0	5	10	15	20	30	40	50	60	70	80
Males	2563	1816	1445	1182	977	686	648	591	489	343	174
Females	4777	3377	2611	2064	1646	1065	886	740	586	409	214

Figure 6. Lifetime attributable risk estimates of all-cancer incidence per 100 000 persons exposed to a single 100-mGy dose to all organs.[21] Cancer risk estimates in this table are for the general US population and may need to be modified if applied to specific subpopulations, such as patients with established coronary disease.

higher in women.[21] Although aspects of these models may be contentious, their underlying idea that cancer risk from radiation is dependent not just on dose but also on nonmodifiable person-specific factors such as age is well agreed on. A thorough discussion of the linear no-threshold model, cancer risk estimation, and their applications to cardiac imaging is beyond the scope of this report, but these subjects remain important areas of investigation.

Conclusions

Effective dose to patients varies widely among standard cardiac imaging studies. *E* for MPI ranges from ≈2 mSv for ^{13}N ammonia and ^{15}O water studies, to ≈10 mSv for standard rest-stress protocols using ^{99m}Tc sestamibi or tetrofosmin, to well over 20 mSv for dual isotope studies. Discrepancies between different data sources are particularly pronounced for ^{201}Tl and ^{82}Rb, and revisitation of the dosimetry of these tracers is warranted. *E* of a 64-slice CTCA scan, with the use of tube current modulation, is comparable to that of a ^{99m}Tc MPI study although somewhat higher in a female patient. CTCA has a lower dose than an MPI study using ^{201}Tl. Dose from CCA varies from 2.3 to 22.7 mSv depending on numerous factors but typically is less than that of MPI or CTCA. For all modalities, careful attention to technique, including the employment of dose-reduction strategies, can minimize dose to patients. Selection of protocols for individual patients and for laboratories needs to be determined from an ALARA approach, and understanding the dosimetry of cardiac imaging protocols is a first step toward implementing a test selection strategy that minimizes risk to patients while providing optimal diagnostic information.

Sources of Funding

This work was supported in part by an NIH/NCRR Clinical and Translational Science Award (1 UL1 RR-24156-01), and by a Nuclear Cardiology Foundation Research Award to Dr Einstein.

Disclosures

Dr Einstein has served as a consultant to GE Healthcare and received travel funding from Philips Medical Systems. Dr Thompson has received a grant from and given lectures for Bristol-Myers Squibb. Dr Moser was formerly employed by Siemens. Dr Cerqueira has given lectures for GE Healthcare, Bristol-Myers Squibb, CV Therapeutics, and Tyco and has served as a consultant to and is a research grant recipient from GE Healthcare and CV Therapeutics. Dr Henzlova has given lectures for Bristol-Myers Squibb and received research grants from GE Healthcare, Molecular Insight Pharmaceuticals, and CV Therapeutics.

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KEY WORDS: angiography ■ radioisotopes ■ computerized tomography, x-ray ■ radiation dosimetry ■ safety