Radiation Protection in Diagnostic Radiology

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Abstract. This paper summarizes technological advances in diagnostic radiography, fluoroscopy and computed tomography, mainly the developments in digital detectors and helical CT scanners with multiple-row detectors. It reviews dosimetric concepts and presents the most up-to-date information on the global patterns of use of radiological modalities as well as the trends in diagnostic medical exposure doses world-wide. Given the increase in population dose, and following the recent recommendations of the International Commission on Radiological Protection, emphasis is placed on the justification of the radiological procedures and the need for referring physicians to consult referral criteria. Protection is optimized by the radiation dose being commensurate with the purpose of the examination. The trade-offs between dose, noise, contrast and resolution are mentioned. The need to establish diagnostic reference levels as an optimization tool and the caveats in comparing some published numerical values are indicated. Methods of dose reduction involving equipment and software design, operational parameters and shielding calculations are discussed, and illustrative numerical examples are given.

KEYWORDS: diagnostic radiology, radiation protection, dosimetry

1. Introduction

Medical imaging has evolved rapidly in the last decades mainly because of the development of new image detectors. Analog systems such as screen/films (S/F) for static images and image intensifiers for dynamic images are being replaced by digital systems such as flat panel detectors. Thanks to the advances in computerization, planar projections are being complemented with 3-D and 4-D imaging. Perhaps the greatest innovations have occurred in computed tomography (CT) where helical (spiral) geometry and multidetector arrays permit to perform a scan of up to 320 slices in a few seconds. Hybrid systems such as SPECT/CT, PET/CT and PET/MR allow the integrated acquisition of dynamic and static images, which can also be obtained by image fusion. The explosion of these technologies and their applications in oncology, cardiology and neurology have resulted in significant increases in the use of diagnostic medical imaging, with the consequential increase in population dose, as the 2008 UNSCEAR Report documents [1].

2. State of the Art - Technological Advances

2.1 Planar Projection Imaging

2.1.1 Radiography/Fluoroscopy

Radiological imaging is a process by which the attenuation of an x-ray beam traversing a part of a human body is either recorded in a medium for later medical interpretation of potential pathology (radiography) or displayed in real-time on a monitor for functional assessment (diagnostic fluoroscopy) or intervention (interventional radiology). New detectors of the computed radiography (CR) or the digital radiology (DR) type are steadily replacing S/F combinations in radiography and image intensifiers and video cameras in fluoroscopy.

As the primary x-ray absorber, S/F radiography uses luminescent intensifying screens, while CR uses storage phosphor screens. In S/F, the emitted light is the only image signal available to the film, the recording medium. Image processing in DR is different. There are two types of DR systems: those that are based on charge-coupled devices (CCDs), and those that are based on amorphous silicon (a-Si:H) flat-panel detectors. Direct DR technologies convert the incident x-ray quanta into a measurable latent image signal, while indirect DR technologies require some intermediate steps. A direct detector produces charge directly from the x-ray absorption within a semiconductor (e.g., a-Se). An indirect

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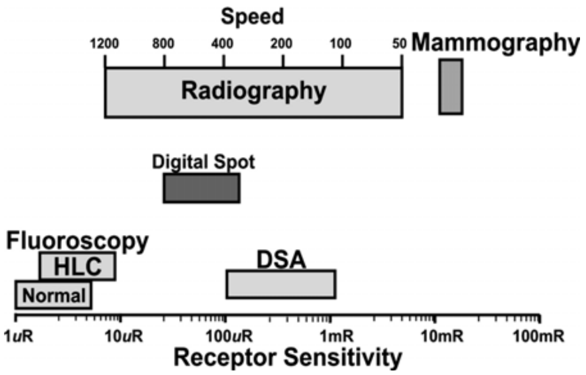
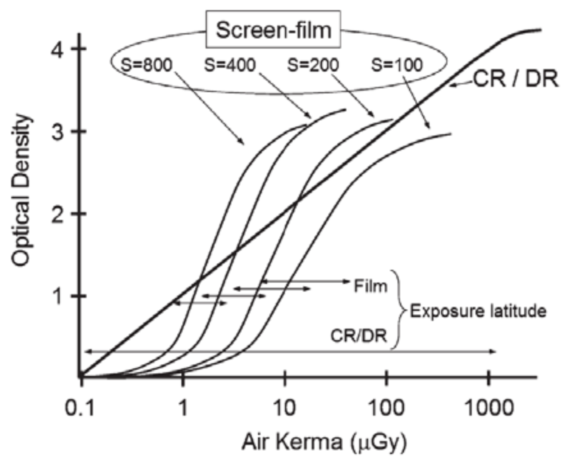
detector produces secondary signals, such as light photons from an x-ray scintillator for subsequent conversion to charge by a thin-film transistor (TFT) photodiode array or charge-coupled device (CCD) camera [2]. Flat panel detectors are transforming both radiographic and fluoroscopic imaging, and are an integral component of image-guided radiotherapy.

The exposure required to form an image varies depending on the sensitivity of the image receptor as shown in Figure 1 from Reference [3]. Other characteristics of the image are contrast and contrast sensitivity (the ability to visualize low-contrast objects), blurring and visibility of detail (spatial resolution), visual noise, artifacts and spatial (geometric) characteristics (magnification and distortion). The major controlling factor for x-ray image noise is the amount of exposure to the receptor. With screen-film radiography, the receptor sensitivity is fixed to a specific value determined by the type of film and intensifying screens used and the quality of the film processing. Because of the relatively narrow film latitude, the exposure to the receptor must be within a limited range, or the films will be either under- or overexposed and appear too light or too dark. Therefore, the noise level in a film radiograph is determined by the design characteristics of the film and the screens that give it a specific sensitivity (speed). On the other hand, digital radiographic receptors have a wide dynamic exposure range, as shown in Figure 2, from Reference [2].

Figure 1: X-ray receptor sensitivity. Receptors for x-ray imaging cover a wide range of sensitivity or speed values. This has an effect on the noise in the images captured by each.
DSA = digital subtraction angiography.
HLC = high level control operating mode
[From Reference 3, pp 25]
(1 mR ~ 10 μGy)

Figure 2. Response of the film to variations in radiation exposure is nonlinear. Contrast enhancement and radiographic speed are based on the characteristics of the intensifying screen (phosphor) and the design of the light-sensitive silver halide emulsion to respond to chemical processing. A selection of screen-film speeds can be chosen to achieve appropriate optical density for a given dose (SNR). CR and DR exhibit a linear response over 10 000 in exposure latitude to allow the user to achieve an SNR sufficient to enable a reliable diagnosis. [From Reference 2, pp 58]





The density of a CR or DR image is entirely arbitrary and is dependent on the processing and completely unrelated to dose. Patient dose can be variable and dependent on the required signal-to-noise ratio (SNR) and efficiency of the detector. Materials used for x-ray detection are broadly classified into three types: (a) turbid unstructured scintillators (e.g., gadolinium oxysulfide [Gd_2O_2S], barium fluorobromide [BaFBr]); (b) structured phosphors (e.g., cesium iodide [CsI], cesium bromide [CsBr]); and (c) semiconductor direct detectors (e.g., amorphous selenium [a-Se]) [2]. There are many types of x-ray converter materials with variations in thickness, absorption, and corresponding resolution. Generally, there is a trade-off between x-ray absorption and resolution; however, with structured phosphors and direct-acquisition detectors, one can achieve both good absorption and good resolution [2].

In an attempt to give the user feedback about the actual detector dose level of a clinical image, most digital systems provide what is called an exposure index (EI). The EI is manufacturer-dependent and it is influenced by the accuracy of the collimation detection algorithm, the collimation, the delay between acquisition and readout, and the reproducibility of the exposure system [2]. It is a good quality control tool but it cannot be translated directly to patient dose. This has to be measured with appropriate dosimeters and a relationship with EIs can be established.

2.2 Volume (3D) Projection Imaging – CT Imaging

The original CT scanner produced images of axial sections by rotating an x-ray tube around the section to be imaged and collecting x-ray transmission data in a bank of detectors that in third generation scanners rotated with the tube. It was a step-and-shoot process with the patient table advancing sequentially after each x-ray tube rotation and creating a set of transmission data, which through a filtered back-projection algorithm, were converted in two-dimensional images of the central plane of each section. Images in planes other than the axial could be generated through multiplanar reconstruction algorithms but their spatial resolution was inadequate clinically. Furthermore, the time to scan large volumes was long, rendering functional imaging practically impossible.

With helical CT, the table and the x-ray tube both move continuously throughout data acquisition, thanks to slip rings, which bring power to the x-ray tube on the rotating gantry, and acquire information from the detector array. The voltage required by the x-ray tube and the transfer signals from the gantry controller are provided by special brushes. [4] The ratio of the table travel per x-ray tube rotation to the collimated x-ray beam width is called the pitch. A pitch of 1.0 indicates contiguous radiation beams; a pitch less than 1.0 indicates overlap of the radiation beams, and a pitch greater than 1.0 indicates gaps between the radiation beams. Another new feature of modern CT scanners is the replacement of each single detector by a row of detectors aligned parallel to the axis of the patient, which has required the broadening of the x-ray beam from a fan beam to a cone beam proportional to the detector group thickness. The data collected from multiple rows can be combined as though they were collected from one detector. The term “channel” has been coined to designate the smallest unit in the z-direction from which data are independently collected. If data from multiple detector rows are collected in such a way that those individual rows are combined, then these rows form a channel. Currently, multi-detector CT (MDCT) scanners are capable of acquiring up to 320 channels of data (along the z-direction) simultaneously [5]. With MDCT, the z-axis detector size is further reduced to submillimeter size, which allows for submillimeter slice thicknesses. Novel detector designs, software advances and high power x-ray tubes permit rotations of less than a second, making volume and functional imaging possible. Additional information can be obtained thanks to dual x-ray energy scanning either by using dual source x-ray tubes or special software. Cardiac applications have benefited from both prospective and retrospective ECG-gated cardiac scanning.

3 Medical Exposure Dosimetry

The magnitudes of equivalent dose and effective dose are radiation protection terms for stochastic effects used by the International Commission on Radiological Protection (ICRP) in prospective dose assessments for planning and optimization in radiological protection, and for demonstration of regulatory compliance with dose limits of occupationally exposed workers and members of the public. Their unit is the sievert (Sv). “Effective dose is not recommended for epidemiological evaluations, nor should it be used for detailed specific retrospective investigations of individual exposure and risk” [6]. In medical exposure, effective dose can be used as a quality control tool to compare doses from different radiation modalities, as shown in Table 1 from AAPM Report 96 [7], but not to assess individual patient doses. The International Commission on Radiation Units and Measurements (ICRU) [8] states that “to assess the risk from stochastic and deterministic effects from medical x-ray imaging, it is necessary to know the organ or tissue doses, the dose distribution and the age and gender of the patients”. The quantities and units to be used in medical x-ray imaging as well as methods for patient dose calculation and measurements are given in ICRU Report 74 [8]. The unit is the gray (Gy).

To determine patient doses in radiography/fluoroscopy, one can measure the kerma in air for all radiation beams and multiply it by the technique factors used, collected either manually or from the DICOM header. Another possibility is to attach to each x-ray machine a properly calibrated air-kerma-area-product meter (previously called DAP meter). CT doses may be calculated from ‘‘computed tomography air-kerma (dose) index” (CTDI) and “air-kerma (dose) - length product” (DLP) measurements [5]. Doses can also be measured directly by placing thermoluminiscent dosimeters or diodes on the patients during the procedures. The dosimetric role of the medical physicist is crucial.

Table 1. Typical effective doses for several common imaging exams from AAPM Report 96 [7]

Radiography/Fluoroscopy (mSv)		CT (mSv)	
Hand radiograph	< 0.1	Head CT	1 – 2
Dental bitewing	< 0.1	Chest CT	5 – 7
Chest radiograph	0.1 – 0.2	Abdomen CT	5 – 7
Mammogram	0.3 – 0.6	Pelvis CT	3 – 4
Lumbar spine radiograph	0.5 – 1.5	Abdomen and pelvis CT	8 – 14
Barium enema exam	3 – 6	Coronary artery calcium CT	1 – 3
Coronary angiogram (diagnostic)	5 – 10	Coronary CT angiography	5 – 15

4 Assessment of Global Practice and Trends in Diagnostic Radiology

According to the 2008 UNSCEAR report [1], approximately 3.6 billion diagnostic (3.1 medical and 0.5 dental) x-ray examinations are undertaken annually in the world. Table 2 shows the estimated annual number of medical x-ray examinations (the only ones to be considered here) and their corresponding annual doses vs health care level [1], an indicator of the number of physicians per million population. (HCL I corresponds to more than 1,000 physicians per million population; HCL II, between 300 and 1,000; HCL III, between 100 and 300, and HCL IV, less than 100).

Table 2. Estimated annual number of medical x-ray examinations and doses, 1997–2007

Health Care Level (HCL)	Population (millions)	Annual frequency per 1,000 people	Annual collective effective dose (10^3 man Sv)	Annual per caput dose (mSv)
I	1,540	1,308	2,900	1.87
II	3,153	332	1,000	0.32
III-IV	1,753	20	33 (III), 24 (IV)	0.03
World	6,446	482	4,000	0.61

There is a significant increase in annual examination frequency for this time period (1997-2007) in comparison with previous UNSCEAR surveys, especially in HCL I and II, where the increase is 60% and 92% respectively with respect to the period 1970–1979. This is mainly due to the ageing of the population, as most medical exposures are performed on older individuals. Regarding dose, as the table shows, the global per caput annual dose from diagnostic medical exposures is 0.61 mSv, which represents 20% of the world total, estimated as 3.1 mSv. (Natural background contributes 79%). This percentage contribution is very different in industrialized countries such as the United States (US) where the rise in medical uses in the period 1980–2006 has resulted in “an increase in the total annual per caput effective dose from 3.0 mSv to 5.4 mSv, making medical exposure equal to or larger than exposure due to natural background” [1]. The figure will increase further if mass screening techniques with CT, currently under clinical trails, such as lung cancer screening for smokers and ex-smokers, calcium scoring for atherosclerosis, coronary angiography and virtual colonoscopy, show benefits.

5 Justification of Procedures (Referral Patterns)

The first step in radiological protection of diagnostic medical exposures is to assess whether the x-ray procedure is really needed, i.e. will the study affect patient management? The *International Basic*

Safety Standards for the Protection against Ionizing Radiation and for the Safety of Radiation Sources (BSS) [9] cautions against “any radiological examination for occupational, legal or health insurance purposes undertaken without reference to clinical indications”. Examples are pre-employment low back x-ray studies, defensive medicine studies to discourage malpractice law suits, and dental x-rays for insurance verification. Another practice that may not be justified is mass screening of population groups “unless the expected advantages… are sufficient to compensate for the economic and social costs, including the radiation detriment” [9]. For example, chest x-ray screening for tuberculosis in areas of the world where tuberculosis is prevalent, and mammography screening in countries with high incidence of breast cancer, are considered justified; whole body CT scans are not. The ICRP [10] listed examples of “low yield examinations”, such as routine preoperative chest radiography, routine prenatal chest radiography, routine pelvimetry in late pregnancy and routine excretory urography in hypertension. Of special concern are mammography examinations in young women without symptoms and who otherwise are at low risk. To provide advice to referring physicians on what medical imaging tests to order, many radiology professional societies have issued guidelines. In the European Union, the medical exposure directive 97/43 [11] requires Member States to ensure that recommendations concerning referral criteria for medical exposure be available to the prescribers of medical exposure. To be used as models, the European Commission (EC) published referral guidelines [12]. In the US, the American College of Radiology maintains updating “Appropriateness Criteria” [13]. WHO, concerned about developing countries recommends that a study be done with only one imaging modality, the one in each health care level that provides the required diagnostic information with minimum cost and dose [14]. Furthermore, before proceeding with a radiological procedure, it is important to ascertain whether there has been any previous examination which would make additional investigation unnecessary [9, 10]. The National Radiological Protection Board (NRPB) of the United Kingdom (UK) –today part of the Health Protection Agency– stated in 1990 that “to eliminate clinically unhelpful examinations” could reduce the annual dose by 20% and to reduce the repeat rate from 10% to 5% by about 4% [15].

Also to be considered, are alternative imaging techniques which do not involve ionizing radiation, such as ultrasound and magnetic resonance imaging (MRI). Typical examples include abdominal, obstetric and gynaecological studies which may be better done with ultrasound and spine and joint studies which may be better done with MRI. Novel techniques such as optical coherence tomography, visible-light, infrared, electrical sources, magnetic sources, impedance, and Terahertz imaging, are being explored [4]. These techniques may be useful when examining pregnant patients and children, who are more sensitive to radiation. For example, the EC Referral Guidelines recommend MRI over CT to examine children suffering from headaches, solely on the basis of radiation exposure [12].

6 Optimization of Protection

6.1 Image Quality / Diagnostic Information vs. Dose

In diagnostic medical exposures, patient dose has to be kept to the minimum necessary to achieve the required diagnostic objective, taking into account norms of acceptable image quality established by appropriate professional bodies and relevant reference levels for medical exposure. This means that exposures resulting in doses above clinically acceptable minimum doses must be avoided. However, it is important to understand that optimization of protection dose not mean dose reduction, and that diagnostic information, not image quality, should be the deciding factor. Optimized images have to be established based on the characteristics of the image receptor, the patient habitus and the purpose of the radiological examination. Particular attention should be paid to children and pregnant females [6].

Since the relationship between exposure and noise is an inverse relationship, noise is the major factor that limits how much x-ray exposure and patient dose can be reduced. This applies to all forms of x-ray imaging, including fluoroscopy and CT. In S/F radiography, the film serves as the acquisition, display, and archival medium and thus, the process must be optimized during the acquisition phase of the image production. In digital radiology, on the other hand, the three processes are separated and image optimization can be accomplished by pre- and post-processing software. Noise reduction methods other than increasing the photon fluence have been published [2].

6.2 Diagnostic Reference Levels

According to the 2007 ICRP [6], "diagnostic reference levels are used in medical diagnosis to indicate whether, in routine conditions, the levels of patient dose… from a specified imaging procedure are unusually high or low for that procedure. If so, a local review should be initiated to determine whether protection has been adequately optimized or whether corrective action is required."

The concept is not new. The BSS [9] called the term "guidance levels" and published values for common diagnostic radiography, mammography, computed tomography and fluoroscopy exams, based on US and UK data. While in the US, the adoption of the diagnostic reference levels (DRLs) is optional; in the EU is obligatory [16]. Wall [17] reported on the European countries that have already implemented the EU Directive. 1999: Spain; 2000: Italy, UK; 2002: Denmark, Finland, Germany, Sweden. The dosimetric parameters chosen differ. Finland, Germany, Spain, and the UK established entrance surface doses (ESD). All the countries except Spain also have DAPs; the UK has DAPs and fluoroscopic time; Germany has ESD and air kerma in addition to DAPs; CT reference levels exist in Denmark, Finland Germany and Sweden. Sweden also has average glandular doses for mammography. The UK revises the reference levels every 5 years. In 2005, it listed DRLs for 30 types of diagnostic x-ray examinations on adults and for 4 types of x-ray examinations on children [18].

In the BSS currently under revision [19], the ICRP terminology has been adopted, and the following requirements are proposed: "such diagnostic reference levels shall be based, as far as possible, on wide-scale surveys that include consideration of image quality, and they are a reasonable indication of doses for average sized patients, and provide guidance on what is achievable with current good practice rather than on what would be considered optimum performance. They shall be applied with flexibility, recognizing that higher exposures may be indicated by sound clinical judgement, and shall be revised as technology and techniques improve, as necessary. In the absence of a wide-scale survey, a set of diagnostic reference levels shall be established by the adoption of appropriate published values". Comparisons between various published reference levels should be done carefully, as both the dose descriptors and the conditions of measurements may differ [20]. For example, entrance surface doses for general radiography in the BSS and in the EC include backscatter factors, the values in the US do not. And while the numerical DRLs values for radiography in the BSS and in the EC are the same, the stated S/F speed in the BSS is 200; in the EC, 400. For mammography, the US and the BSS use the average glandular dose for a compressed 4.5 cm breast; the EC uses the entrance dose for a 5 cm breast. For CT, the BSS adopted the term multiple-scan average dose (MSAD), the EC used instead CTDI and DLP, a practice now universally adopted.

7 Methods of Dose Reduction (Patients, Workers, Public)

Basically there are two methods of patient dose reduction; those associated with the equipment and software and those involving the selection of imaging techniques by the operators. Any dose reduction on patients will also diminish the dose received by the occupationally exposed staff and the public. For the latter two, additional protection is provided by structural and ancillary shielding.

7.1 Equipment Design and Software Applications

The main criterion for selection of radiological equipment should be the clinical objective of the examination. Not to be used are: For mammography, conventional x-ray units; for interventional procedures, fluoroscopes designed for gastrointestinal exams, and, for general radiographic work in fixed installations, portable or mobile x-ray units. The dose reduction could be 100 % or higher.

For S/F technology, dose will be reduced depending on the S/F combination selected, as shown in Figures 1 and 2. (Direct film without intensifying screens is never acceptable). Rare-earth screens, when used with green-sensitive film, may decrease radiation doses by as much as 50%. Screen speeds should be carefully chosen depending on how much resolution (detail) is needed. Film processing, especially replenisher and water flow as well as developer temperature, are critical for dose reduction.

In regards to digital detectors, UNSCEAR [1] has documented cases of dose savings when replacing CR systems with flat panel direct DR. "For the same image quality, radiation doses were halved using direct DR during excretory urography". "Doses were 2.7 times lower for a direct DR compared with S/F radiography for chest imaging and by 1.7 times compared with a CR system". "Effective doses from direct DR for standard radiographic examinations in an accident and emergency department were typically 29% and 43% lower than for S/F or CR". In all the reported cases of full digital mammography, the mean glandular doses were either comparable to those for S/F systems or lower.

Regarding conventional fluoroscopy, since, with a properly operating image intensifier, radiation dose rates to the patient can be reduced to about one-third of those in direct fluoroscopy [10], direct viewing fluoroscopes should be replaced by either fluoroscopes with image intensification or with flat panel detectors which can reduce the dose further. Dose savings are possible with pulsed fluoroscopy (where dose reduction can amount to 70% [21]), last image hold, virtual collimation and wedge filters.

Automatic exposure control (AEC) was first used in S/F radiography. Phototimers terminate the exposure when it is estimated that the film has received the required density. Systems for film radiography are set to give similar optical densities at each tube potential. However, the variation in sensitivity of digital detectors with photon energy is significantly different. Whereas the response of a rare-earth film-screen system will remain relatively constant over the range 80 kV - 120 kV, that of a CR system (barium, fluor-halide) will decline by 10% between 80 and 100 kV and a further 10% by 120 kV. CR systems have dose detector indices which are related to the amount of light generated from the phosphor and setting the AEC cut-off level to achieve similar dose detector index values at each tube potential, should ensure similar levels of image quality across the whole kV range [22].

Fluoroscopic units also have automatic exposure (brightness) controls. To keep the brightness of the TV monitor at a stable level, regardless of what the part of the anatomy is being visualized, it is necessary to keep the dose rate at the input of the image intensifier relatively constant. To do this, the tube voltage and the tube current and/or pulse width must be adjusted as the radiation is transmitted trough the patient. The AEC system in current fluoroscopy units will adjust the tube current (mA) and tube voltage (kV) according to pre-determined programs. A standard anti-isowatt curve in which both mA and kV are raised together will be used for a range of standard examinations. Another option in which the kV is raised more rapidly will provide lower dose and still be satisfactory for applications such as paediatric fluoroscopy and barium examinations. Other options in which the kV is maintained at a lower level will provide higher contrast images but at a higher dose [22].

Regarding CT, manufacturers have reduced doses not only by implementing noise reduction algorithms, but also by increasing the x-ray beam filtration, improving the x-ray beam collimation and the detector geometric efficiency and by developing automatic exposure control (AEC) schemes [7]. To adjust the mAs to compensate for different levels of attenuation of the scanner's x-ray beam, one could vary the tube current or the rotation time. Since variations in rotation time are restricted by the need to perform most scans within a limited time frame, such as a single patient breath hold, the parameter that is varied is the mA. There are three types of AEC in CT scanning [5, 7]. The AEC system adjusts the tube current based upon the overall size of the patient: in this case, the same mA is used for an entire examination or scan series. To take into account the variation of the attenuation along the patient's z-axis (along the scanner couch), the tube current can be adjusted for each rotation of the x-ray tube. In rotational AEC, the tube current is decreased and increased rapidly (modulated) during the course of each rotation to compensate for differences in attenuation between lateral (left-right) and A-P (anterior-posterior) projections. In general, lateral projections are more attenuating than A-P, particularly in asymmetric regions of the body, such as the shoulders or pelvis. The amplitude of mA modulation during rotational AEC reflects the patient asymmetry, with less modulation occurring in regions where the patient is more circular, such as in the head. More modulation occurs in asymmetric regions. The optimal modulation for the AP views has been reported to use less than 10% of the mA used through the lateral views, when measured in shoulder phantoms [23].

Image noise is affected by rotational AEC in a different way from patient and z-axis AEC. The effect of rotational AEC on the image is to even out variations in image noise across the field of view. This also reduces the severity of photon starvation artifacts through asymmetric body regions, and thus it offers the most clear cut dose reduction of CT AEC systems. ECG-controlled tube current modulation has also proven to be very effective in dose reduction in MDCT scanning of the heart. Dose reductions of 48% for males and 45% for females, respectively have been reported [24].

All these equipment-based dose reduction methods have resulted in lowering the doses for single examinations. The use of x-ray imaging equipment that delivers the lowest reasonable practicable dose to the patient compatible with the clinical outcome of the examination is a statutory requirement in the UK [21], following EU directives [11]. Decisions regarding changes to be made in existing equipment should be subjected to a cost-effectiveness analysis, as recommended by a UK Working Party [21].

7.2 Operational Parameters (Number and Types of Exams and Selection of Techniques)

The critical parameters in regards to dose reduction are the number and type of projections in each static examination and the duration of the exposure in dynamic imaging. The NRPB [15] estimated that to limit the number of radiographs per examination could reduce the dose by 20% and choosing projections that could minimize dose to sensitive organs, by 50%. In digital radiology the speed of the exam not only induces the acquisition of many more images than needed but also allows the unnecessary retake of diagnostic images considered of substandard quality. Furthermore, the ability to improve image quality at the post-acquisition stage makes the operator less concerned about choosing the technique that will give a lower dose for the same image quality. Yet the same physical parameters that control analog systems control digital ones. In radiography/fluoroscopy, dose depends on tube potential and filtration, exposure technique (mAs), patient x-ray tube distance, patient image-receptor distance, collimation (field size), and image processing factors. In general, x-ray studies should be done at the highest tube potential (kV) and the lowest tube current-product (mAs) to produce an image with minimum acceptable contrast and maximum acceptable noise that yields the required diagnostic information. As the voltage and filtration are increased, the radiation doses to the parts of the patient in the beam are generally reduced for a given exit dose because of the increased penetrating power of the radiation. To compensate for the concomitant loss in the radiographic contrast of bone, anti-scatter grids are used. The NRPB estimated that the choice of grids with lowest grid factors compatible with adequate scatter rejection could result in a 20% dose reduction factor and removing the grids during fluoroscopy when detail is not critical by 50% [15]. Modern radiographic systems have preselected choices for typical radiographic procedures and patient sizes. A 50% dose reduction can be achieved if the exposure factors are matched to patient habitus [15].

In fluoroscopy, when two dose rate modes are available, the high dose-rate mode should only be used when absolutely necessary for the clinical situation. For systems with image intensifiers and AEC, the main dose savings available to the operator is the choice of field size, not just by adjusting the diaphragm to the minimum necessary to cover the area of interest, but mainly by selecting the image intensifier field size. Since the gain of the image intensifier is lower at the smaller field size, the AEC system increases the kV and the mA to achieve the required brightness on the monitor. When changing from a 23 cm to a 15 cm and from a 15 cm to an 11 cm field of view, the corresponding entrance exposure increases by factors of 2.3 and 4.0, respectively. When changing from a 23 cm to an 11 cm field of view, the entrance kerma rate increases by a factor of nine [25]! In addition, decreasing the tube current and the time in fluoroscopic examinations, the dose could be reduced by 30% [15].

In CT, dose depends, in addition, on scan time, scanner rotation angle, angular position of the x-ray tube at the start of the exposure, patient position within the field, patient orientation, slice thickness, spacing and pitch. For complete rotations (360^0) and non-pulsed x-ray beams, dose is directly proportional to scan time in the axial scan mode and to rotation time in the helical mode. The fastest rotation time should typically be used to minimize motion burring and artefact, and the lowest tube potential consistent with the patient size should be selected to maximize image contrast [7]. Modern CT systems have a range of pre-programmed protocols for different examination types, with set values for tube potential, current, rotation time, slice thickness, etc. The techniques may be adjusted to each institution’s needs at the time of acceptance testing, when dose is to be measured. Special attention should be paid to paediatric CT scans.

When specifying an imaging protocol in CT, it is very important to note the detector configuration used to acquire the desired slice thickness, as this significantly affects the subsequent retrospective reconstruction options (for thinner or thicker images) and the radiation efficiency of the system (i.e., patient dose). For instance, using an MDCT scanner one might acquire 5-mm slices either by using a wide beam collimation (4 x 5 mm) or a narrow one (4 x 1.25mm). The wide beam collimation allows much faster *z*-coverage, while the slower narrow beam collimation acquisition allows retrospective reconstruction of narrower slice thicknesses, albeit with relative radiation dose inefficiency [7]. The ratio of the multiscan to the single scan dose depends on the slice thickness, the slice separation, the numbers of scans taken and the shape of the single-scan distribution. When acquiring data in the spiral mode, all CT scanners require an additional rotation or so of data collection at the beginning and end of the scan in order to obtain sufficient data to reconstruct images over the prescribed volume. The percentage of "wasted" x-ray beam increases as slice thickness decreases but diminishes when there are more detector rows, since there are fewer contiguous overlapping beams in the same volume. This means that for the same scan length, 64-MDCT deliver less dose than 4-MDCT [5]!

7.3 Shielding Considerations

In addition to the shielding that the x-ray unit assembly itself provides to the parts of the body which are not to be imaged, some patient organs like gonads, breast, thyroid and eyes within or adjacent to the primary x-ray beam can be shielded with leaded-impregnated materials placed over them and whenever possible, surrounding them. Occupationally exposed personnel should always wear protective leaded aprons when performing fluoroscopy.

Workers and public will mainly be protected by structural shielding and restricting access to control areas. The dose constraints used in shielding calculations are really part of the optimization process and therefore should take into account economic and societal factors. The concern is that when using too low a constraint, many countries (mostly developing ones) are spending unjustified amounts of money that curtail the funds available for health care. Rodgers [26] estimated that in the United States the added cost of shielding to 0.25 mSv/year would be 20% for new construction of diagnostic rooms, and the total cost for retrofitting existing diagnostic rooms, a minimum of US$2 billion. The recent US National Council on Radiation Protection and Measurements (NCRP) [27] Report on barrier thickness calculations for various x-ray rooms uses a shielding design goal of 1 mSv/year for public exposure.

8 Conclusions

Mainly because of the increase in the frequency of x-ray examinations, patient doses have increased in the last few decades. The results of many studies in several countries are documented in UNSCEAR 2008 [1]. This Report also analyzed occupational doses among six identified subgroups in diagnostic radiology: CT technologists, general radiographers, fluoroscopy technologists, radiologists, nurses and radiologic technology interns. It concluded that more than 80% of CT technologists and general radiographers do not have measurable exposure and that the average individual effective dose for interventional procedures is significantly higher than for conventional diagnostic radiology [1].

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