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*Review*

# Patient dose reduction methods in computerized tomography procedures: A review

**M. A. AWEDA and R. A. AROGUNDADE**

Department of Radiation Biology, Radiotheray and Radiodiagnosis, College of Medicine/Lagos University Teaching Hospital, P. M. B. 12003, Lagos, Nigeria.

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**Health hazards are associated with medical exposures to ionizing radiation, including during Computed Tomography (CT) procedures. Rapidly increasing number of CT facilities worldwide is accompanied by enhanced staff, patient and public radiation doses and these needs to be controlled to minimize health risks. This paper reviews the relevant patient and CT scanning parameters that influence patient dose during the common diagnostic procedures. These include scanning geometry, tube current, applied high potential (kVp), scanning mode and length, collimation, couch speed and pitch, gantry rotation speed and radiation shielding. The paper also presents some strategies for limiting patient dose through modulation of exposure parameters and design of technical devices for image processing. These include collimation, filtration, automated modulation of tube current, use of adaptive reconstruction and noise filters. Patient weight and size of the scanned anatomical part influence the absorbed dose, x-ray beam collimation and filtration can reduce dose by 17% to 50%, tube current modulation can lead to 10% to 60% reduction, projection adaptive reconstruction filter can reduce dose by 30% to 60% while noise filters can produce 17% reduction in noise variance compared with the conventional filters.**

**Key phrases**: Computerized Tomography, Patient Dose Reduction; Scanning Parameters Modulation.

## INTRODUCTION

Computerized Tomography (CT), like other imaging modalities using ionizing radiation, has in recent years experienced tremendous technological advances, developing from the first generation in the early 1970s through seventh generation (Bushberg et al., 2002) to Multi-dimensional CT (MDCT) (Mori et al., 2005a and b; Mori et al., 2006) . Compared with other imaging modalities using x-rays, radiation doses from CT are relatively high according to ICRP (Rehani et al., 2000) and often approach or even exceed the values known to increase the probability of cancer formation (Gray, 1996; Brenner et al., 2001). A review of the literature revealed a rapid global increase in the frequency of CT procedures; hence increase in radiation dose to the population, staff and patients. The United Nations Scientific Committee on the Effects of Atomic Radiation report on Medical Radiation Exposures (UNSCEAR, 2000) stated that CT constituted only 5% of radiological examinations, but contributed about 34% of the collective dose In the UK, CT contribution to the collective effective dose from medical exposures in 1999 was 40%, compared with 20% in 1990 (Crawley et al. 2001). CT accounts for about 11% of x-ray based medical procedures in the USA, but delivers over 67% of total dose associated with medical imaging procedures (Mettler et al., 2000). Over 600,000 abdominal and head scans are performed annually in children below 15 years, while about 500 of those scanned may ultimately die of cancer attributable to CT radiation exposures (Brenner et al., 2001).

It is obvious that the benefits derived from diagnostic radiation exposures exceed the harmful effects of the radiation exposure. The increasing use of CT facilities increases radiation doses to the staff and the population, and this call for continuous efforts in dose reduction. Various methods and strategies based on individual patient attributes may be devised for this purpose. Failure of dose reduction efforts may lead to a reversal of the risk-benefit ratio associated with this imaging modality. The risks associated with CT procedures may be deter-

*Corresponding author. E-mail: awedama@yahoo.com. Tel: +2348032501228, Fax: + 234 01-5851432

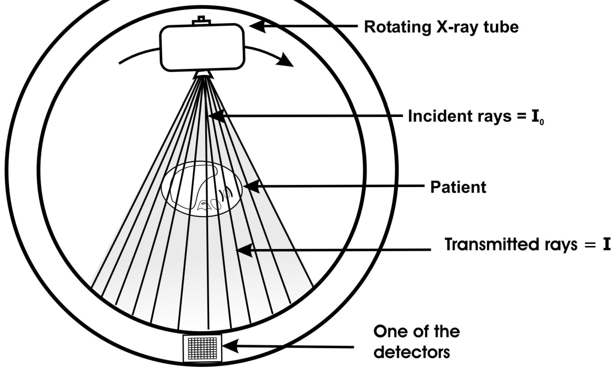


**Figure 1.** X-ray tube with a complete circular array of detectors.

ministic due to cell death and which is quantified in terms of the dose to a particular region with a threshold level beyond which these effects occur. The second type of risks is stochastic, where the probability of occurrence depends on the absorbed dose. Optimization of patient dose is the only way to avoid or minimize these effects, while still achieving satisfactory image quality (IAEA, 2004). This paper outlines the basic principles of CT procedures with emphasis on patient dose reduction methods through the manipulation of some appropriate scanning parameters.

## Principles of CT imaging and dosimetry consideration

CT is a unique imaging modality with a set of specific parameters that influence patient dose. It is also unique because during examinations, exposure is continuous and it is around the patient. Typical examinations use multiple exposures along some length of the patient in order to cover the entire volume of the region to be examined. The CT comprises of three major components, viz Gantry, Radiation Source and Detector. The radiation beam incident on the patient traverses the body. During the passage the incident beam intensity $I_o$ is attenuated according to:

$$I = I_o e^{-\mu x} \text{ ------------------------------------ } (1)$$

In a proportion that depends on the local tissue composition (average atomic number Z), density ρ and the patient thickness x. I is the emergent intensity and μ is the linear attenuation coefficient that depends on Z and ρ. i.e. in soft tissues attenuation is less while in bones, it is more. The attenuated beam emerges from the patient to reach the detectors which through interactions convert electromagnetic radiation to electrical signals. The detectors are in a complete circular array in which the x-ray tube rotates with the divergent fan-shaped x-ray beam (Figure 1). The signals so generated are used for image reconstruction. The energy of the x-ray beam which depends on the applied kVp to the tube, the photon fluence which depends on the anode current and exposure time are some important parameters that influence patient dose (Rehani and Berry, 2000). As the beam traverses the patient, energy deposition is rather uniform across the plane scanned since the patient is equally irradiated from all directions, unlike in conventional radiography where the dose decreases continuously with depth from the skin entrance to the exit of the beam. In the CT of the head, dose is distributed uniformly across the field of view. In large objects as in abdominal scans, it is uniformly distributed around the periphery of the patient and decreases slowly towards the center.

**Figure 2.** Typical phantom used for determination of CTDI.

Radiation energy deposition in tissue in CT procedures is not limited to the scanned volume alone. Adjacent tissues as well absorb radiation due to the scattering, the divergence of the beam and limited efficiency of the collimator. The conventional parameter representing the integrated dose within and beyond the scanned volume is the Computerized Tomography Dose Index (CTDI) measured in mGy (Bushberg et al., 2002). The CTDI is equivalent to the Multiple Scan Average Dose (MSAD) that results from a series of scans spaced by the nominal section thickness. CTDI is defined as the integrated dose profile (in the z-direction) for a single slice, normalized to the nominal slice thickness. It can be measured either in air or in a phantom using either a pencil ion chamber or a row of TLDs. In essence the CTDI gives a measure of the raw output of a scanner. Figure 2 shows typical phantoms for determining CTDI and for CT quality control procedures. In Figure 3 is a 10 cm ionization chamber used with the phantom for measurement.

Common forms of CTDI include $CTDI_{air}$, measured at the centre of rotation of the beam in the absence of patient or phantom (without scatter and attenuation) and CTDI measured at the centre of a PMMA phantom (16 cm of diameter for head scans and 32 cm for body scans)

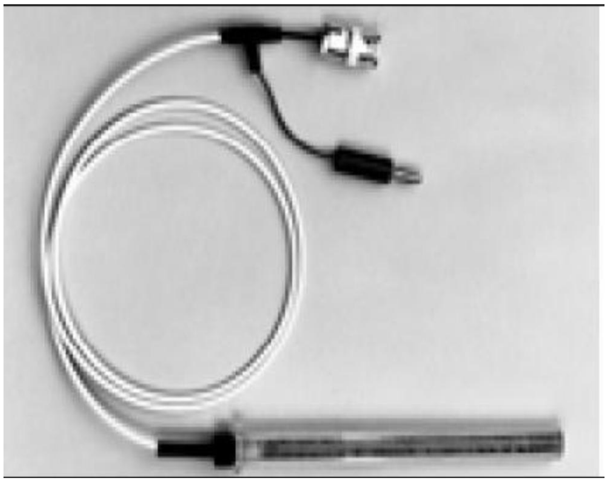

**Figure 3.** 10 cm ionization chamber used with the phantom for determination of CTDI.

as $CTDI_c$, and at the phantom periphery (1 cm depth) as $CTDI_p$.The definition is as follows:

$$CTDI_w = 1/3CTDI_c + 2/3CDTI_p \text{ ---------- } (2)$$

where N is the number of slices, T (mm) is the nominal slice thickness.

Another CTDI descriptor that takes into account the parameters related to a specific imaging protocol, the helical pitch or axial scan spacing is defined as

$$CTDI_{vol} = CTDI_w \times NT/I \text{ -------------------- } (3)$$

with NT/I = 1/pitch. The factors that influence CTDI have been identified as kVp, mAs, pitch and collimation. The influence of each factor according to McNitt-Gray (2002) is presented in Table 1 below.

Patient organ doses and thus effective dose are calculated using the NRPB Monte Carlo dose data for CT scanners (Jones and Shrimpton, 1993). Both the CTDI and Dose length parameters are displayed on the monitor in some modern CT systems. Image quality is largely influenced by noise and it is inversely related to the radiation energy. A decrease in tube current or tube voltage results in dose reduction but causes an increase in noise. This implies that dose reduction is a crucial issue in the sense that while attempting to minimize patient dose image quality should not be compromised, hence in practice; optimal parameters have to be chosen (Rehani et al., 2000). Efforts on dose reduction must therefore be considered along with image quality and standard practice. The challenge to the Medical Physicists and the Radiologists is therefore to identify the acceptable threshold of image quality with minimum possible patient dose in conformity with the As Low As Reasonably Achievable (ALARA) principle.

**Relevant scanning parameters affecting patient dose**

The scan geometry, the tube current, the applied high tension, the scanning mode, the length of the scan, the speed of the couch, the speed of rotation of the gantry and shielding are some major parameters that influence patient dose in CT procedures. The operator can monitor most of these parameters and modify them to achieve the desired image quality with minimum possible patient dose.

**Scanning Geometry:** The distance between the focal spot of the tube and the isocenter depends on the geometry. A single- or multiple-detector row helical CT can have a long or short geometric configuration. The intensity of the radiation beam varies between the source and the patient according to the inverse square law of distance. Therefore with all the other scanning parameters fixed, a short geometry scanner will produce more interactions in, therefore more dose to the patient and will have lower image noise than a long one.

**Tube Current:** Reducing the tube current or the beam intensity is a way of reducing patient dose. A 50% reduction in tube current reduces dose by half. The beam energy and the photon fluence vary with the kVp and the tube current in a given procedure. The current-time settings (mAs) are proportional to the photon fluence. Although some authors have claimed that it is possible to reduce the current without adverse effect on image quality (Sohaib et al., 2001; Kalra et al., 2002; Donnelly et al., 2001), such reduction should be made with caution because it is accompanied by increase in image noise, which degrades image quality. This is particularly so in abdominal scans where low contrast regions are greatly affected by noise.

**Tube Potential (kVp):** This determines the radiation quality and its variation causes variation in patient dose. The relationship between kVp and image quality is complex because it affects both the image noise and the tissue contrast. Decrease in kVp causes increase in noise. This is particularly so when the patient size is large and the current is not appropriately increased to compensate for the low kVp. Dose is proportional to the square of the change in kVp while the latter is inversely proportional to the noise change. The choice of high tension is therefore crucial. An optimal kVp for abdominal scan for an averagely sized patient may be 120 kVp instead of 140 kVp, this will lead to 20 to 40% reduction in patient dose (Kopp et al., 2002). This value has to increase for a large size patient for adequate beam penetration. Lieberman et al. (2002) have published the results of a study showing that skull CT in children at substantially reduced tube potential with increased tube current produced the lowest possible dose without compromising the contrast-to-noise ratio and image quality.

**Table 1.** showing the Variation of CTDI with kV, mAs, Pitch and Beam Collimation.

| Variation of CTDI with kV | Energy (keV) | CTDI for Head Scan (mGy) | CTDI for Abdominal Scan (mGy) |
|---|---|---|---|
| | 80 | 14 | 5.8 |
| | 100 | 26 | 11 |
| | 120 | 40 | 18 |
| | 140 | 55 | 25 |
| **Variation of CTDI with Current** | **(mAs)** | | |
| | 100 | 13 | 5.7 |
| | 200 | 26 | 12 |
| | 300 | 40 | 18 |
| | 400 | 53 | 23 |
| **Variation of CTDI with Current** | **Pitch** | | |
| | 0.50 | 80 | 36 |
| | 0.75 | 53 | 24 |
| | 1.00 | 40 | 18 |
| | 1.50 | 27 | 12 |
| | 2.00 | 20 | 9 |
| **Variation of CTDI With Single Detector Collimation** | **Collimation (mm)** | | |
| | 1 | 45 | 19 |
| | 3 | 41 | 18 |
| | 5 | 40 | 18 |
| | 7 | 40 | 18 |
| | 10 | 40 | 18 |
| **Variation of CTDI With Multidetector Collimation** | **Collimation (mm)** | | |
| | 5 | 62 | 33 |
| | 10 | 46 | 24 |
| | 20 | 40 | 20 |

**Scanning Mode:** Use of a multi-detector row CT scanner results in some part of the beam extending beyond the edges of the imaging region. This is because at the beginning of data acquisition, only the first detector row contributes to imaging (Toth, 2002). As the acquisition proceeds, additional detector rows enter the imaging region until all the rows contribute. As a result, it is generally more dose-efficient to use a single helical scan rather than multiple helical scans if there are no overriding clinical considerations such as breath holding of the patient. The need to prescribe multiple contiguous helical scans should be infrequent with modern high speed multi-detector row scanners.

**Scanning length:** With the increasing availability of helical CT scanners today, there is a tendency to extend the area of coverage to include regions beyond the actual area of interest in the chest, abdomen, or pelvis, which will further increase patient dose. Therefore, it is essential to draw the attention of referring physicians and radiologists to consider the consequence on the patient dose and to establish scanning protocols that restrict the examination to what is absolutely essential.

**Collimation, Couch Speed, and Pitch:** In helical scanners, pitch is defined as the ratio of couch feed per $360^o$ gantry rotation to the normal collimator width of the x-ray beam. An increase in the pitch decreases the duration of exposure of the patient being scanned, hence the patient dose. Beam collimation, couch speed, and pitch are interlinked parameters that affect the image quality. Faster couch speed for a given collimation resulting in higher pitch will reduce patient dose, especially if other scanning parameters, including the tube current, are kept constant. This is because of the shorter exposure time, whereas narrow collimation with slow couch speed results in a longer exposure time, and hence higher patient dose. This is not true for scanners that use effective milliampere-second (mAs) setting and maintain a constant mAs value. In such scanners, the effective mAs value is held constant irrespective of pitch value, so that the dose does not vary when the pitch changes. For a given collimation, an increase in couch speed increases the pitch and reduces the radiation dose (Rehani et al., 2000, McNitt-Gray et al., 1999).

Although scanning at a higher pitch is generally more dose efficient, it tends to cause helical artifacts, degradation of the section-sensitivity profile or section broadening, and consequently, decrease in spatial resolution. Alternations in pitch can have varying effects on image quality and in different situations. For instance, in CT colonoscopy, (Bogoni et al., 2005), image quality and reconstruction artifacts are less affected by pitch than by beam collimation, so that a higher pitch with narrow beam collimation are preferable for reducing dose (Laghi et al., 2003). However, in some situations such as metastatic liver, which generally require thin collimation, an increased pitch may affect the detectability as the lesion may be missed owing to degradation of the section-sensitivity profile.

Due to “overbeaming” in multi-detector row CT, some amount of the x-ray beam is incident beyond the edges of the detector rows (Kopp et al., 2002, McCollough; Zink, 1999). Generally, thicker beam collimation in multi-detector row CT results in more dose-efficient examination, because overbeaming constitutes a smaller proportion of the detected x-ray beam. Depending on the scanner type, thick collimation limits the width of the thinnest sections that can be reconstructed. On the other hand, although thin collimation increases the proportion of overbeaming, it allows reconstruction of thinner sections. Hence, beam collimation and pitch must be carefully selected to address specific clinical requirements. For instance, a thicker collimation and a pitch greater than 1:1 is usually sufficient for screening in colonography and the urinary tract calculus. However scanning in certain clinical situations such as liver resection or transplantation, work-up is often performed with thin collimation and a pitch of less than 1:1.

**Gantry Rotation Time:** There has been a dramatic decrease in the tube rotation times with recent technological innovations, most notably with development of 4-, 8- and 16-detector row CT scanners. A 4-row scanner with 0.8 s rotation time requires 16 s breath hold to scan the entire abdomen, while an 8-row scanner will cover this length in 8 s. If the tube rotation time is decreased the exposure time will decrease and the tube current may have to be increased to maintain constant image quality. Modern 16-row scanners are capable of high scanning speeds and sub-millimeter section thickness. Thin collimation can lead to higher dose, especially if tube current is increased to maintain image noise at a level similar to that of thicker sections. The contrast resolution of small lesions improves because of reduced partial volume effect; hence greater noise on thinner sections may often be acceptable (Hu and Fox, 1996). In addition, submillimeter collimation scans can normally be reconstructed as thicker sections, which reduce inherent noise. Thus it is important to optimize beam collimation for different multi-detector row scanners.

**Shielding:** Protection of radiosensitive organs such as the breast, eye lenses and gonads, is particularly relevant in paediatric patients and young adults, because these structures frequently lie in the beam pathways. Beaconsfield et al. (1998) have reported that with lead shield, thyroid and breast doses were reduced by 45 and 76% respectively in 110 procedures. Therefore shielding of the tissues not included in the examination is helpful in reducing patient dose. If the gonads are included in the field but are not the organs examined, some form of shielding could be used. Hidajat et al. (1996) have again reported reduction in dose to the testes up to 95%, using testis capsule during abdominal CT procedures, whereas lead apron is not appropriate for the ovaries due to their non-constant position. Hein et al. (2002) as well reported the use of shield for protection of the eye lens in paranasal sinus CT as a suitable and effective means of reducing patient dose by 40%.

**Anatomical parameter consideration in dose reduction**

Most patient dose optimization methods involve modulation of the scanning parameters, especially tube current, on the basis of patient weight and cross-sectional abdominal size. That is to say that weight and patient size also influence patient dose.

**Weight:** Several investigators have suggested that mAs value can be substantially reduced for CT of the chest in both adult and children (Prasad et al., 2002, Diederich et al., 1999). Image quality identical to that in adult can be obtained in paediatric patients using signif-cantly reduced exposures. For abdominal CT, Donnelly et al. (2001) described modulation of scanning parameters in children on the basis of weight. They reported that pae-diatric patient weight can be used to select appropriate mAs that are much lower than for adult in abdominal CT. They also suggested the use of substantially reduced mAs for children weighing 4.5 to 68.0 kg. For abdominal CT in adults, tube current can be reduced on the basis of patient weight (Kalra et al., 2002). Selection of CT parameters on the basis of a patient’s weight can lead to large variation in image quality between, for instance, two persons with the same weight but different heights.

**Cross-sectional Dimension:** Attenuation of the incident x-ray beam in CT depends on the size of the body portion being studied; and greater exposure is required in corpulent patients to attain image quality equal to that in slimmer patients. For the same exposure needed to compensate for a large size patient, the image quality is better with a slimmer patient because more photons reach the detector and the image noise is reduced. However, the dose to the slimmer patient is higher than necessary to produce good diagnostic image. Scanning parameters can, therefore, be modified on the basis of cross-sectional sizes to optimize patient dose. Haaga et al.

(1981) have reported that image noise was related to patient cross-sectional area and advocated the use of cross sectional measurements for optimizing scanning parameters and dose. A new method recently reported is patient dose variation in order to achieve similar levels of image noise for patients with different abdominal diameters (Starck et al., 2002). Modulation of scanning parameters using anatomical diameter has yielded a dose reduction in slim patients and a significant correla-tion has been reported (Kalra et al., 2002) between patient dose reduction, image quality and abdominal cross-sectional parameters such as abdominal circumfe-rence, cross-sectional area, and anteroposterior and transverse diameters. At 50% reduced tube current (i.e. about 50% of the patient dose), image quality was accep-table in patients with a cross-sectional area of less than 800 $cm^2$, a circumference of less than 105 cm, a root mean square diameter of less than 44 cm, an anteropos-terior diameter of less than 28 cm and a transverse diameter of less than 34.5 cm. Conversely, image quality with reduced tube current was unacceptable in patients with larger abdominal dimensions (i.e., exceeding the aforementioned values). These dimensions can be estimated before examination with a caliper. Alternatively, the technologist or the Medical Physicist can directly measure them on the CT console monitor. McCollough et al. (2002) evaluated the use of size-based CT charts for reducing dose to paediatric and small patients and for improving image quality in large patients. They reported that modification of tube current in proportion to patient width is feasible and that it results in a 2- to 4-fold dose reduction in small patients.

## Technical parameter consideration for dose reduction

A variety of technical strategies that aim at decreasing patient dose in CT procedures have been developed, and many others are still in experimental stage. The majority of the technical innovations address patient dose optimization by improving scanning efficiency and image quality thus aiding image acquisition with reduced exposure. These innovations include collimation of x–ray beams, use of better filters and image processing algorithms, automatic tube current modulation, and efficient detector configuration and shielding.

**Beam Collimation:** Focal spot tracking, control of x-ray tube focal spot motion, and beam collimation enhance scanning efficiency. Overbeaming is reduced by measuring the beam position every few milliseconds and continual repositioning of the source aperture so that a narrow beam reaches the detector. The beam is thus stabilized on the detectors, with exposure profile narrower than the detected x-ray profile, and the patient dose associated with multi-detector row is reduced in comparison to that of systems with no focal spot tracking.

**Beam Filtration:** X-ray filters absorb the soft x-rays that constitute superfluous radiation which do not reach the detectors and thus do not contribute to image formation, but contribute to patient dose. Efficient filters selectively remove soft x-rays to reduce patient dose. Itoh et al. (2001) compared doses with a 5.8 mmAl with conventional filter in a phantom and patient study. They noted a 17% reduction in dose and a 9% decrease in image noise with the new filter. Bow-tie filters and beam-shaping filters reduce the skin dose by 50% compared with flat filters (Toth T. L., 2002). Bow-tie and beam-shaping filters minimize dose in the thinner proportion of patient, thereby providing better noise consistency within the image while saving substantial amount of radiation exposure.

**Automated Tube Current Modulation:** In the CARE Dose system, during each rotation of the tube and detector assembly around the patient, a small number of the central detector channels provide attenuation information, which is dependent upon the patient cross section and scan angle, to the X-ray generating system (Kalender et al., 1999). The information provided by these detector channels is used to determine to what extent the mA can be modulated, with respect to an initial tube current setting, without adversely affecting the image quality. As a result the tube current is modulated dynamically with a delay of one rotation relative to the attenuation measurement. The first patient based assessment by Greess et al. (2000) showed that, when CARE Dose is used, a dose reduction of approximately 25% (in terms of total mAs reduction) is possible in pelvic scanning "with no significant decrease" in subjective assessments of image quality. Similar percentage dose reductions have been demonstrated in other clinical work (Greess et al., 2002) and these showed good agreement with phantom based data (ICRP 1996, Kalender et al., 1999). Most of the published work has used image noise and/or subjective image assessment to quantify image quality. A small number of papers, Mastora et al. (2001) and Jacobs et al. (2002) have used standard deviations from regions of interest (ROIs) to yield a more objective assessment of image noise. Tube current modulation is a new technical innovation that can substantially reduce dose (Iball et al., 2006). The concept of automatic tube current modulation is based on the premise that pixel noise is attributable to quantum noise in the projections. By adjusting the tube current to follow the changing patient anatomy, quantum noise can be adjusted to maintain the desired noise level.

There are two current modulation methods used in CT scanners today: the longitudinal (z-axis) and angular (x- and y-axis) modulation. In z-axis modulation, tube current is adjusted to maintain a user-selected quantum noise level. Noise is regulated on the final image to a level desired by the user. Z-axis modulation is the CT equivalence of the automatic exposure control systems used for many years with the conventional x-ray systems. It is an attempt to make all images have similar noise, independent of patient size and anatomy. The dose savings in z-axis modulation are expected to be greater than those with fixed-tube current methods since the tube current will be automatically reduced for smaller patients and anatomical regions. Z-axis modulation has been recently introduced for multi-detector row CT scanners such as Autom A by GE Medical Systems. Tube current modulation is determined from the attenuation and shape of scout scan projections in the patient just prior to the CT examination. Clinical results of these techniques have not yet been published in the literature.

In angular modulation, the tube current is adjusted to minimize x-rays projections that are of less importance for the reduction of the overall image noise. In anatomical parts that are highly asymmetric such as the shoulders, x-rays are much less attenuated in the anteroposterior direction than in the lateral direction (Greess et al., 2000). Thus, the overwhelming abundance of anteroposterior x-rays can often be reduced greatly without a marked effect on the image noise. Angular modulation was first introduced on single-detector row scanners in 1994 (Kalender et al., 1999). Dose reduction of up to 25% was reported at that time, with virtually no change in image noise. On these early systems, both lateral and anteroposterior scout scans were required to determine angular modulation. More recently, angular tube current modu-lation has been introduced on multi-detector row scann-ers (CARE Dose by Siemens, Erlangen, Germany). In this system, the modulation is determined in real time by using projection data that lag by 180° from the x-ray generation angle. A recent investigation of 100 helical CT imaging studies in children in which angular modulations were used showed a 10 to 60% decrease in dose, with a mean reduction of 22.3% (neck, 20%; thorax, 23%; abdomen, 22 %) without loss of image quality (Greess et al., 2002).

The ideal CT scanner will employ both z-axis and angular modulation techniques. When available in all commercial CT scanners, use of manual techniques, in which a tube current value is selected on the basis of some simple measurements on the patient (e.g. weight or cross-sectional dimensions), will be replaced with this computerized objective approach. With these developments, tube current modulation in CT scanners will be similar to photographic timing or automatic brightness controls like those currently used in conventional radiography. Indeed, automatic tube current modulation promises to be an important development in the optimization of scanning parameters that will help eliminate the guesswork involved in exposure parameters selection.

**Projection-Adaptive Reconstruction Filters:** A marked decrease in signal is common in regions such as the shoulders due to beam attenuation in a particular projection. This leads to increased image noise and reduction in image quality that result from photon contamination by the electronic noise of the data-acquisition system. Projection space filters increase the filtration of signal-dependent noise in the reconstruction data and thus minimize the loss of resolution. Although there is some loss of image resolution accompanying the use of these filters, this is less than 5%, and the use of projection-adaptive reconstruction filters prevents an otherwise diagnostically compromised image. Kachelriess et al. (2001) investigated the use of multi-dimensional generalized adaptive filters for reducing image noise and patient dose. They recorded 30 to 60% reduction in image noise, typically along the direction of the highest attenuation in the non-cylindrical body regions such as shoulders and metallic implants, without an increase in radiation dose.

**Noise Filters:** As discussed earlier, patient dose reduction is limited by increased image noise that can obscure lesions otherwise visible with standard parameters. Noise-reduction filters have been designed to decrease image noise and patient dose. Alvarez and Stonestrom (1979) reported that two-dimensional linear filtering of the image may alter the spatial resolution and noise of CT images. They developed filters that minimized the variation in noise subject to a constraint on spatial resolution, with a 17% reduction in noise variance in comparison with that of conventional filters. Use of nonlinear image-processing techniques for improved quality CT images obtained with lower doses has also been reported (Keselbrener et al., 1992). Recently, Yu et al (2002) reported the use of a new algorithm for reconstruction of CT images with noise properties superior to those of image reconstruction with the conventional fan-beam filtered-back projection (FFBP) algorithm currently used in commercial CT systems, including multi-detector row scanners. This algorithm converts the fan-beam data to non-uniformly sampled parallel-beam data using the Fourier shift theorem in the angular direction. The approach performs ramp filtration on non-uniform sampling grids along the radial direction before back projecting the filtered data to form the image. The decrease in noise with this algorithm may be translated into reduced patient dose and enhanced detection of subtle lesions, compared with reconstruction based on the current widely used FFBP algorithm.

Noise reducing filters have also been designed on the basis of the principle that a group of structural pixels representative of structures of interest and a group of non-structural pixels representative of non-structural regions are both present in any image (Kalra et al., 2003a, b). The structural pixels can be identified by determining gradient values for each pixel and by identifying pixels with a desired relationship to the gradient threshold value. The noise reducing filter technique involves isotropic filtering of non-structural regions with a low-pass filter and directional filtering of structural regions with a smoothing filter operating parallel to edges and an enhancing filter operating perpendicular to the edges. A blending parameter regulates the recombination of the structural and non structural segments. Noise-reducing filters decrease noise on low-dose CT images but adversely affect contrast and sharpness and may therefore decrease lesion contrast (Kalra et al. 2003a, Kalra et al., 2002). Further improvement in the technique is needed to maintain image contrast while decreasing image noise.

## Conclusion

One of the fundamental principles of radiation protection requires that requests for CT procedure must be prescribed exclusively by qualified medical practitioners and must be justified by both the referring physician and the radiologist. Establishment of clinical guidelines to advise the referring physicians and the radiologists on the appropriateness and acceptability of CT examinations will help minimize superfluous patient exposure. In addition, CT procedures should not be repeated without clinical justifycation (Rehani et al., 2000, Rehani; Berry 2000). Procedures with non-ionizing radiation such as ultrasonography and magnetic resonance imaging should be considered as alternatives for appropriate clinical indications when equal or greater diagnostic information could be obtained. For instance one of the benign conditions responsible for the largest cumulative patient dose from CT is complicated acute pancreatitis, for which it is possible to substitute MRI for CT.

CT images are often acquired before, during and after intravenous administration of contrast materials. When possible, multiple exposures should be reduced by eliminating precontrast imaging. This is possible in the evaluation of liver and bowel wall conditions, where precontrast images can often be omitted without affecting the results. As recommended by the International Commission on Radiological Protection, all CT performed for research purposes but without immediate benefits to the individuals undergoing the examination should be subjected to critical evaluation, since the absorbed doses could be quite higher than those of conventional radiography. A critical step towards uniform optimization of CT radiation dose is the establishment of standard protocols for all examinations on the basis of patient size, weight and the scanning features such as imaging noise and automatic modulation of tube current. This will ensure that good quality images are acquired with patient doses that are reduced to the lowest possible levels.

CT patient dose optimization is a crucial issue that must be addressed by the Physicists, the Radiologists and manufacturers of CT scanners. The benefits of precise diagnosis to the patient should always be greater than radiation risks. Radiologists in conjunction with the Medical Physicists should adopt consistent strategies for limiting patient radiation dose, while manufacturers should focus efforts towards improving CT technology necessary for diagnostic image quality with reduced radiation dose. Concerted efforts and research should be directed towards defining and achieving high image quality, technology-based methods and modulation of the relevant parameters to achieve a diagnostic quality CT image at a minimal dose.

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