A Technique of Computer-Simulated Dose Reduction for Conventional Chest Computed Tomography (CT).

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Abstract: Computed Tomography is a diagnostic imaging modality giving higher patient dose in comparison with other radiological procedures. The level of CT radiation dose is of concern to radiologists, medical physicists, government regulators, and the media. This review addresses this problem with particular reference to radiation dose in chest CT. Specifically it outlines the topics of measurement units used to quantify radiation exposure, factors affecting CT scanner dose efficiency, scanner settings that determine the administered radiation dose, and radiation dose reduction in chest CT**.**

The Objective: Aim of this study is to determine minimal tube current (mAs) required for good image quality on conventional chest CT (Computed Tomography).

Patients & Methods: Prospectively, 35 consecutive patients (mean weight, 65 kg; range, 38-92 kg) older than 45 undergoing conventional chest CT with standard technique (120 kVp, 400 mAs) had four additional sections imaged at reduced tube current (200, 140, 80, 20 mAs) at two levels (tracheal carina and left atrium). CT scans were evaluated in random order by two independent observers who were blinded to technical factors used.

Results: The 400 mAs scan was considered the reference standard. When compared with the reference technique (400 mAs), the first and second (200 mAs and 140 mAs) reduction levels showed no significant difference **($p > .05$)** in subjective image quality. A significant difference (**$p < .001$)** was seen at the third and fourth (80 mAs and 20 mAs) reduction levels. A two fold reduction in tube current (400-140 mAs) and resultant radiation dose did not cause a significant change in subjective image quality or in detection of lung abnormalities with conventional chest CT. One hundred forty milliampere-seconds is the minimal tube current required to provide good image quality in patients of average weight.CT is a diagnostic imaging modality giving higher patient dose in comparison with other radiological procedures. CT scan parameters have been adjusted with the aim of working towards optimization of image quality and patient dose.

Conclusions: The effective dose of our population is (1.8 mSv) compared with UK, Crawly et al and Hughes et al have the same value.

Key words: Radiation dose, CT scan& Reduction dose.

Introduction:

The introduction of helical or spiral, computed tomography (CT) in the late 1980s revolutionized diagnostic medical imaging Single–detector row CT scanners and, more recently, multi–detector row CT scanners markedly increased the clinical indications for CT [1]. CT plays a major role in chest disease investigation. However, concern has been raised about radiation doses [2, 3].To our knowledge, tube currents have been chosen arbitrarily without assessing impact on image quality and lesion detectability. Appropriate tube current is more difficult to define for CT than for conventional radiography because CT is a digital technique in which acquisition and display are not related. Therefore, when tube current is excessive, the CT image does not become too dark but merely improves because of decreased image noise.Because radiation dose is linearly related to amperage at afixed kilovoftage, reduction in the milliamperage on tube current used is equivalent dose reduction. Thus, optimal CT tube current is an appropriate balance between image quality and radiation dose [4, 5].

The increase in population radiation exposure from CT, particularly in children, has been of concern to radiologists, medical physicists, government regulators, and the media [6]. The suggestion that excessive radiation doses are being prescribed for CT has appropriately aroused the attention of the radiologic community [7, 8]. Radiologists and medical physicists must be attentive to their responsibility to maintain an appropriate balance between diagnostic image quality and radiation dose [9].

The simplest parameter, radiation exposure, is determined by measuring ionization in air caused by the x-ray beam. The measurement unit is coulombs per kilogram (abbreviation, C/kg). It has limited clinical value, as it does not take into account the area irradiated, the penetrating power of the radiation, or the radiation sensitivity of the irradiated organs. From radiation exposure we can calculate the skin entrance dose, which is important when examining deterministic effects such as skin erythema. Although deterministic effects are not encountered in routine CT, they are of potential concern in CT fluoroscopy [10]**.**

The concept of reduced tube current for conventional 10-mm-collimation chest CT was introduced by Naidich et al. [7] in 1990 with demonstration of acceptable image quality for assessment of lung panenchyma with low milliamperage settings (20 mAs). These images were adequate for assessing lung parenchyma, but they had considerable increase in noise [11].

Most centers use **120 kVp,** but there is no consensus about optimal tube current. Tube currents from 200 to 533 mAs for chest CT have been reported [12]**. To our knowl**edge, tube currents have been chosen arbitrarily without assessing impact on image quality and lesion detectability. Appropriate tube current is more difficult to define for CT than for conventional radiography because CT is a digital technique in which acquisition and display are not related. Therefore, when tube current is excessive, the CT image does not become too dark but merely improves because of decreased image noise.

Because radiation dose is linearly related to amperage at a fixed kilovoftage, reduction in the milliamperage on tube current used is equivalent to dose reduction. Thus, optimal CT tube current is an appropriate balance between image quality and radiation dose [13, 14].

Subjects & Methods:

CT collective effective dose can be reduced in several ways, the most useful strategies being justification of each individual examination by a consultant radiologist, reduction of the scanned volume, optimum selection of technique factors (kV, mA, rotation time, slice width and pitch (for helical scans) or couch increment (axial scans)). All these strategies were employed in the process of optimizing patient dose and image quality for CT scan protocols at this hospital following the installation of a helical CT scanner. The original scan protocols used were based on those recommended by the manufacturers as a starting point for clinical work, but these were not optimized as the optimization process requires local clinical input.

A study was done on 35 consecutive patients (18 women, 17 men) older than 47 years (range, 49-86 years; median, 72 years) who had conventional 10-mm-collimation CT chest scans. Average subject weight was 68 **± 14** kg (mean ± **SD; range,** 34-93 kg). **IV contrast material** was used in 16 cases. Subjects initially had chest CT with contiguous 10-mm sections and conventional technical parameters (120 kVp, 200 mA, 2-sec scan) used at our institution. This 400-mAs scan will be referred to as the reference technique.

At completion of this diagnostic study, four additional 1 0-mm sections were imaged at two levels (carina and inferior pulmonary veins) using decreasing amperage settings (200, 140, 80, and 20 mAs). These levels were chosen to allow assessment of the mediastinum (tracheal carina level) and to evaluate effect of increased soft tissue thickness and cardiac motion (level of inferior pulmonary veins). All imaging at reduced tube current used a kilovoltage and scan time identical to those used for the reference scan.

The CTDOSE software requires the following input parameters: scanned volume (in terms of baseline in the phantom and number of slices), slice width, couch increment (axials), effective mAs and CT dose index per mAs (CTDI). CTDI is a measure of the total dose from a single slice irradiation. The European working document gives the following formula for CTDI [15]:

$$\boxed{CTDI = \frac{1}{T}\int_{-\infty}^{\infty} D(Z)\,dz}$$

The quantity dose–length product (DLP) was then derived for all scan protocols using the methods described in the European working document, for comparison against the four proposed diagnostic reference levels relating to head, chest, abdomen and pelvis [15]. The quantity DLP uses a weighted CTDI, ($CTDI_w$ (mGy)). $CTDI_w$ is an approximation to the average dose over a single slice and is derived from a combination of measurements at the surface and centre of a defined set of Perspex phantoms, according to the equation:

$$\boxed{CTDIw = 1/3\ CTDIc + 2/3\ CTDIp}$$

Where $CTDI_c$ is the CTDI measured at the centre of the phantom and $CTDI_p$ is the CTDI measured at the periphery of the phantom.

The European working document gives the following formulae for DLP:

$$\boxed{CTDI_{VOL} = \frac{CTDI_W}{Pitch}}$$

Where pitch is defined as table distance traveled in one 360˚ rotation / total collimated width of the x-ray beam.

$$\boxed{DLP = CTDI_{vol}\ x\ Scan\ Length}$$

The definition of DLP is Therefore, DLP increases with an increase in total scan length or with the variable affects $CTDI_w$ (e.g. tube voltage or tube current) or the pitch .Because scan length is expressed in centimeters, the SI unit for DLP is (mGy .cm).

The effective dose reflects the non uniform radiation absorption of partial body exposure relative to a whole body radiation dose and allows comparisons of risk among different CT examination protocols. A reasonable approximation of the effective dose can be obtained using the equation:

$$\boxed{\mathbf{E = E_{DLP}.\ DLP}}$$

Where E is the effective dose and E_{DLP} is a conversion factor **($mSv.mGy^{-1}.cm^{-1}$)** that varies dependent on the body region that is imaged

Results:

A linear relationship was found between subjective image quality and square root of milliamperage from 20 to 200 mAs **with and without IV contrast medium**. However, this linear relationship did not hold for the 400-mAs scan. Repeated measures analysis of variance showed no significant difference ($p > .05$) in subjective image quality between reference tube current (400 mAs) and the first or second (200 mAs or 140 mAs) reduction levels (Figs. 1). However, a significant difference ($p < .001$) in subjective image quality was seen at the third and fourth (80 mAs and 20 mAs) reduction levels.

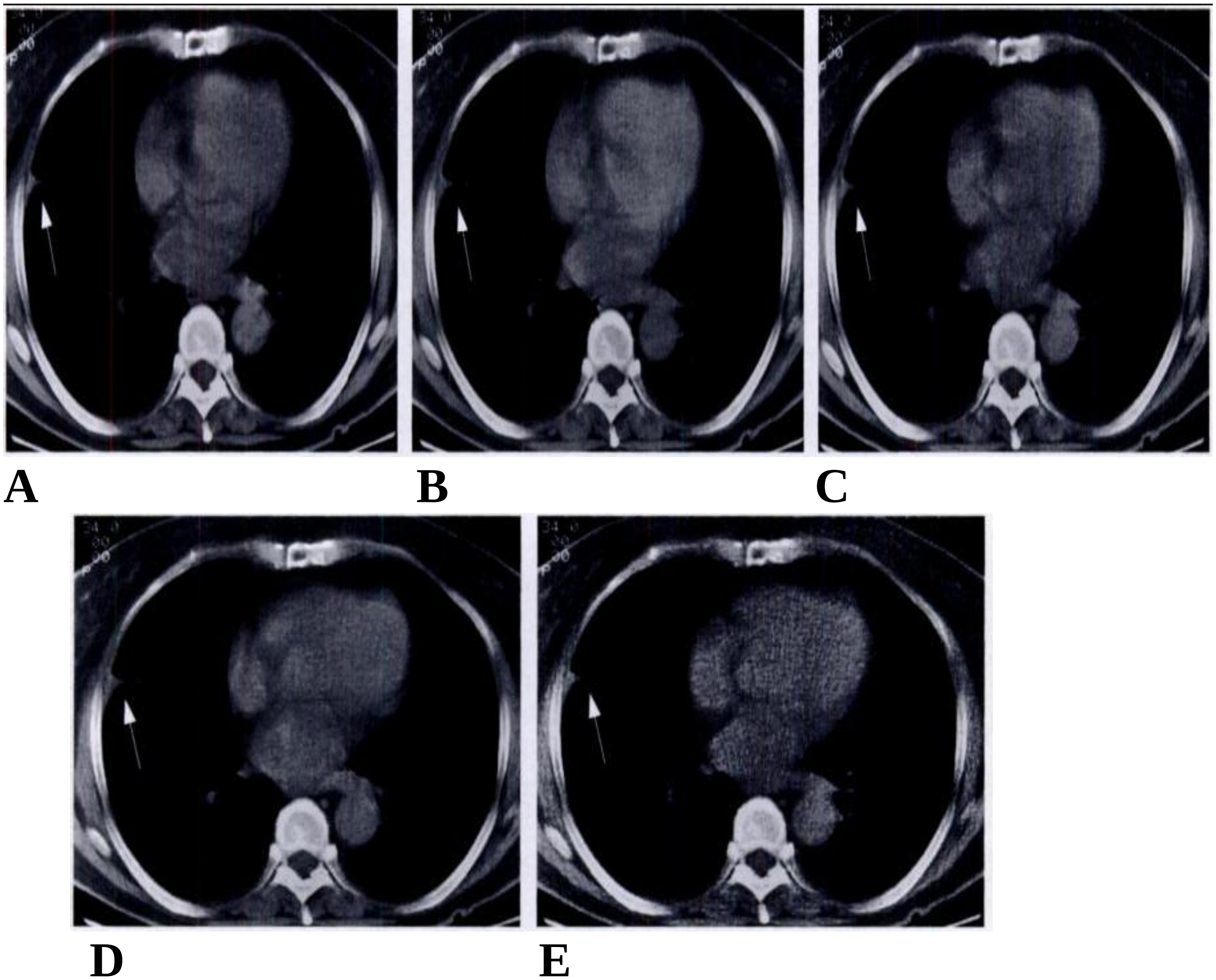

Fig. 1- Chest CT scans at:
A, 400 **mAs.** ***C,*** 140 **mAs.**
B, 200 **mAs**. *E,* **20 mAs** D, 80 mAs

Effective doses for each scan protocol are shown in **Table 1**, together with

The collective effective doses in each type of chest examination, before and after dose reduction process.

The **CTDI$_w$** and DLP values used for the scanner to calculate effective dose in this study are given in **Table (2&3)**.

Table (1): Effective dose before and after dose reduction

Examination	Numbers	Type of chest examination	Before dose reduction (mSv)	After dose reduction (mSv)
Chest	35	**Standard dose**	3.9	1.8
		Chest cancer	4.8	2.1
		Chest for lymphoma	3.9	1.6
		High resolution chest	0.87	0.87
		Oesophageal cancer	3.3	1.4
		Average chest	3.4	1.4

Table (2): The Weighted CTDI Calculating by the CT unites based on the measured (CTDIcenter) and (CTDIperphery), kVp=120 and mAs=140

CTDIw =1/3 CTDIc+2/3 CTDIp				
Type of examination	**kV**	**CTDIair mGy**	**CTDIw mGy**	**CTDIair ICRU mGy**
Chest	120	0.334	0.158	0.357

Table (3): Comparison of highest dose–length products (DLPs) against proposed reference levels

Examination	Type of chest examination	DLP (mGy cm)	Proposed reference level (mGy cm)
Chest	**Chest for cancer**	146	643

Discussion:

As a result of the CT technique, recognition of excessive radiation dose is more difficult than in plain radiography. However, excessive radiation dose has been used if an insignificant change in subjective image quality or lesion detection occurs with increasing milliamperage.

In this study, higher subjective image quality scores were consistently obtained with higher tube current. However, incremental gain in subjective image quality at higher tube current was less than at lower tube current. Subjective image quality improved an average of 47% from 20-140 mAs, where as quality improved only 17% from 140-400 mAs. A linear relationship was identified between subjective image quality and square root of tube current from 20 to 200 mAs. This supports the concept that image noise was the major determinant of subjective image quality for this range of mAs and is consistent with results of previous studies [16, 17]. No significant change in subjective image quality was seen between 200-mAs and 400-mAs scans, suggesting that for these milliamperage levels, image noise reduction did not cause a detectable improvement in subjective image quality.

Significant difference in subjective image quality between upper or lower section levels suggests that radiation dose is more important than cardiac motion for determining image quality. Because of reduced image smoothing [18].

The effective dose per examination, averaged across all scan protocols, has returned to about (1.8 mSv) the same level as **Crawley et al** (1.8 mSv) 1991[19] and Hughes et al (1.7 mSv) 1999 [20].

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