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Dose and image quality in CT

Introduction

For any CT examination, an explicit choice must be made for the X-ray tube voltage (kVp) and X-ray tube output (mAs). The choice of X-ray tube voltage and output will affect the quality of the CT image as well as the radiation dose received by the patient [1, 2, 3]. In this paper, the manner in which patient dose and image quality depend on the X-ray tube voltage and output is described. Dose and image quality in CT are closely linked. It is generally possible to increase image quality at the cost of a higher dose. It is also possible to reduce the dose at the cost of lower image quality.

Radiation in CT

Biological effects

Radiation effects are either deterministic or stochastic. Deterministic effects include skin erythema, epilation, and the induction of cataracts. Deterministic effects are characterized by a threshold dose which must be exceeded before the effect will occur. For example, the threshold dose for the induction of cataracts is of the order of 5 Gy for protracted (chronic) exposures. In general, doses in CT are well below deterministic threshold doses, and one does not expect to see effects such as erythema, epilation, and cataracts.

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Stochastic effects (the random effect of radiation not dependent on a threshold) include carcinogenesis and the induction of genetic mutations. It is the cancer risk that is deemed most important. For most CT patients, the principal radiation risk is later malignancy. Knowledge of cancer risks is obtained from epidemiological studies of groups exposed medically, occupationally, or in the A-bomb exposure in Hiroshima and Nagasaki in 1945 [4, 5].

To quantify the risk to patients undergoing CT examination, we need a measurable quantity that is related to the patient risk. However, the normal dose quantities in radiology of exposure and absorbed dose are "concentration units", and do not reveal the total amount of radiation absorbed by a patient. This point may be illustrated from chest X-ray examinations. One patient has a small area exposed (e.g., 10×10 cm) with an entrance skin dose of 0.1 mGy, and a second patient has a large area exposed (37×45 cm) with the same skin dose of 0.1 mGy. In these two cases, the skin dose is exactly the same because the film blackening needs to be the same. However, the risk to the patient with the large exposure area will be nearly 20 times as large, because the total energy imparted is 20 times as high. Energy imparted is proportional to the area exposed if the skin dose is constant.

Measuring radiation

The computed tomography dose index (CTDI) is a "concentration unit" and does not directly quantify the patient risk. Consider a single section obtained during a CT examination. The CTDI corresponds to the total energy deposited in the patient or phantom, divided by the mass of that section. In effect, the CTDI is the average dose in a directly irradiated region when a large number of contiguous sections are acquired. The patient risk is only indirectly related to the CTDI. To illustrate this point, compare the CTDI values for a 1-mm CT section and a 10-mm CT section irradiated using the same X-ray tube voltage and output. The CTDI values will be very similar, but the energy (and therefore the risk) imparted to the 1-mm section is about one-tenth that for the 10-mm section.

The patient risk in CT depends on the dose to all organs and tissues and on their relative radiosensitivity [6, 7, 8]. Organs with a high radiosensitivity include the red bone marrow (leukemia risk), colon, lung and stomach. The effective dose (E) can be obtained by combining the dose to each organ with the radiosensitivity of that organ, and then summing over all the exposed organs and tissues exposed during a CT examination.

The effective dose is the best single parameter for quantifying how much radiation a patient receives during any radiological examination [9, 10, 11]. Effective doses can be converted into a numerical risk of dying from a radiation-induced cancer. The International Commission on Radiological Protection (ICRP) recommends the use of a risk factor of 5% per Sievert, deemed appropriate for a general population which includes children [12]. However, risk factors need to take into account the demographic features of exposed populations and are subject to considerable uncertainty at the dose levels encountered in diagnostic radiology.

CT doses

CTDI

The amount of radiation absorbed by a patient depends on the size and composition of the patient as well as the choice of radiographic factors (X-ray tube voltage and output) used in a CT examination. In this section, the importance of scanner settings and patient characteristics for determining doses in head CT examinations is discussed.

Most of the increase in cranial size occurs in the first 2 years of life. The composition of the head also increases from about 50 Hounsfield units (HU) for newborns to about 200 HU for adults. For radiation dosimetry purposes, heads may be modeled as equivalent cylinders of water so that the mass of a CT section of a patient is equal to the mass of the equivalent cylinder of water. Infants can be modeled as a 60-mm radius cylinder of water, whereas the corresponding radius for adults is 90 mm.

Figure 1 shows how the CTDI for a single section (340 mAs) varies with X-ray tube voltage and patient size. The CTDI increases significantly with increasing X-ray tube voltage. CTDI values in infants are higher than in adults. The reasons for the higher doses in infants are their smaller size and the lower amount of attenuation of the primary X-ray beam.

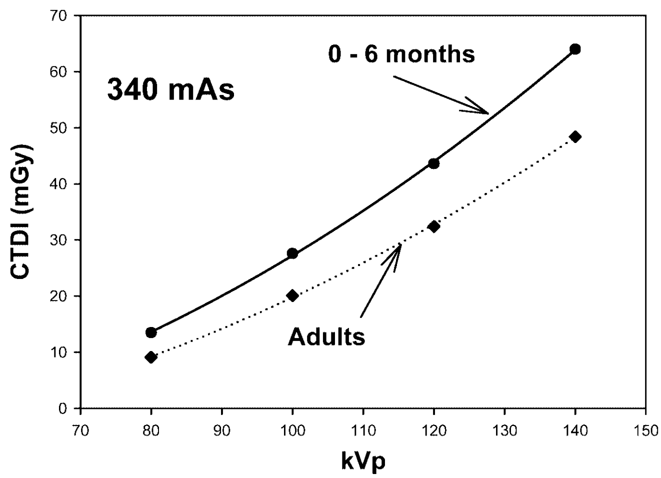

Fig. 1. CTDI versus X-ray tube voltage at a constant 340 mAs

Energy imparted

The CTDI data shown in Fig. 1 do not represent the total radiation received by patients because they take no account of section thickness or of the total number of sections scanned. For illustrative purposes, we will assume that a patient receives 18 7-mm sections. An important and useful parameter that describes the total radiation received by the patient is the energy imparted. Energy imparted is directly proportional to both the section thickness and the total number of sections acquired.

The energy imparted for a *single* section is the product of the CTDI and the mass of the directly irradiated section (i.e., $\pi r^2 T$) where r is the radius of the water equivalent cylinder that models the patient head and T is the section thickness. The *total* energy imparted to the patient is the product of the energy per section and the total number of sections. It is easy to understand whether the energy imparted increases or decreases as scan parameters are modified. Doubling the number of CT sections, for example, will generally double the energy imparted.

Figure 2 shows how energy imparted is affected by patient size and X-ray tube voltage. Although adults have low CTDI values (Fig. 1), they absorb more energy than infants because they are bigger. The energy imparted can easily be computed; it can also be converted into the corresponding effective dose.

Effective dose

For a given X-ray procedure of a specified body region, doubling the energy imparted will double the radiation risk. Knowledge of the dose distribution for a single CT scan can provide both the effective dose (E) and the energy

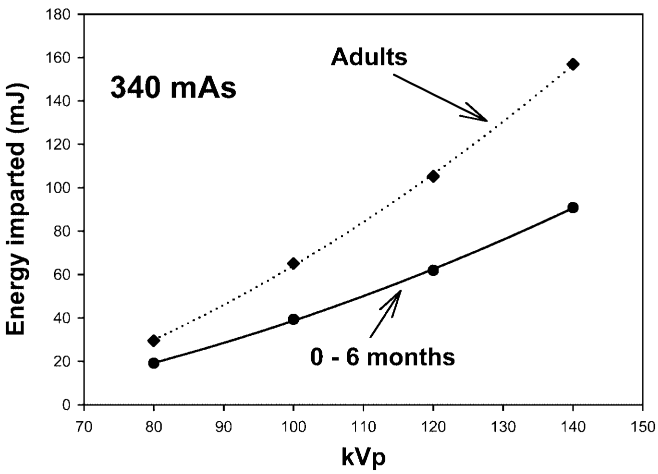

Fig. 2. Energy imparted versus X-ray tube voltage for patients having a head CT examination with 18×7-mm sections (340 mAs)

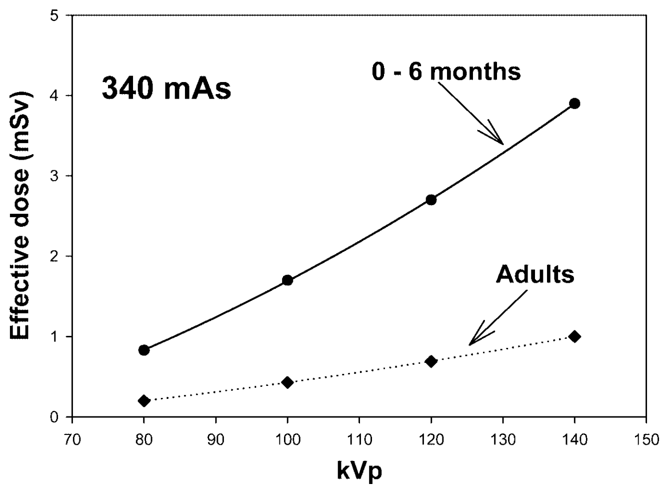

Fig. 3. Effective dose versus X-ray tube voltage for patients having a head CT examination with 18×7-mm sections (340 mAs)

imparted (ϵ), and thereby permit the generation of the ratio of E/ϵ [13]. An independent estimate of energy imparted (see above) can then be used to generate the effective dose to an adult patient undergoing any type of head CT scan. Similar calculations can be performed on infants and children to provide patient size-specific E/ϵ ratios [14].

Calculations of E/ϵ ratios have been performed using Monte Carlo dose distribution data. Adult head CT examinations have E/ϵ ratios of the order of 5 mSv/J, but the corresponding E/ϵ ratios in infants are higher. In effect, these ratios show that although children absorb less radiation than adults, their organs are so small that the resultant doses (energy absorbed divided by organ mass) are higher. When scanned by the same techniques, one would therefore expect effective doses in children to be higher than in adults [15, 16].

Figure 3 shows effective dose values that correspond to the energy imparted data shown in Fig. 2. Note that the patient effective dose increases by about a factor of five when the X-ray tube voltage increases from 80 to 140 kVp. Also note that the effective doses in infants are much higher than in adults. Furthermore, children are more radiosensitive than adults, and the risk to an infant who receives 1 mSv is much higher than the risk to an adult from the same radiation dose [17].

Image quality

Contrast

In diagnostic radiology, image contrast is related to photon energy but is generally independent of X-ray beam intensity (i.e., mAs). The average photon energy is primarily dependent on the X-ray tube voltage, but will also depend on the filtration and the size of the patient. Of great importance is the type of lesion being investigated. High atomic number (Z) lesions (e.g., those containing an iodinated contrast agent) are expected to have a high contrast relative to water, whereas low Z lesions (e.g., soft tissue) will have a low intrinsic contrast.

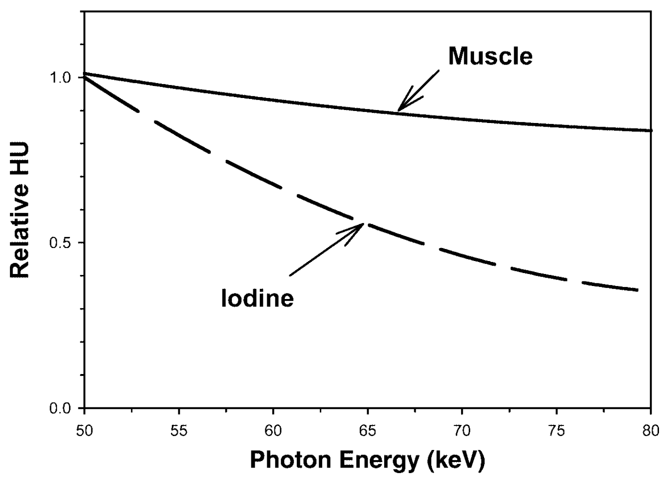

Fig. 4. Relative HU versus photon energy for the specified lesion in a water background

Figure 4 shows how the relative contrast of both high Z and low Z lesions, as expressed by their relative Hounsfield unit values, varies with photon energy [18]. Increasing the photon energy markedly reduces image contrast for an iodine lesion; the corresponding drop in image contrast for a low Z (muscle) lesion is much lower.

Noise/mottle

Image noise (mottle) is the random fluctuation in intensity for the same nominal radiation exposure. In CT, quantum mottle is the dominant source of image noise and is solely dependent on the "number of photons" used to create the image. Increasing the X-ray tube output by 100% will reduce image noise by 41.4% since quantum mottle is proportional to the square root of the number of photons used to create the image. Note that if the X-ray tube voltage is increased, this will reduce image contrast (see above) and also reduce image noise since there are more photons produced at higher X-ray tube voltages.

Mottle is important because it limits the visibility of low contrast lesions. Figure 5 shows empirical data for the noise level in liver CT scans acquired using 120 kVp and 400 mAs [19]. The data in Fig. 5 show that as patient size increases, the reduced detected X-ray intensity causes the measured level of image noise to increase. Use of a constant X-ray tube output results in the image quality being determined by the size of the patient. Clearly the X-ray tube output should be adjusted to ensure that the level of noise in the CT image is determined by the diagnostic task at hand [20, 21].

Contrast to noise ratio (CNR)

Image contrast depends on photon energy. Image noise depends on the number of photons used to make the image. However, noise is essentially irrelevant if the level of contrast is high, whereas noise is of paramount importance if the intrinsic contrast is low. In practice, the ratio of contrast to noise (CNR) is the primary determinant of CT image quality. If the CNR level is high, lesion detection will be easy. Low CNR will limit lesion detection and could be improved either by increasing the contrast or by reducing the noise.

Figure 6 shows the measured level of noise in a uniform water phantom as a function of X-ray tube output [19]. At a given output, the 32-cm diameter phantom has a higher level of noise than the 16-cm phantom because of the greater attenuation of X-rays in the larger phantom. These curves illustrate that to maintain a constant noise level, scanning smaller patients requires much lower X-ray tube output values. Another interesting feature of the data presented in Fig. 6 is that the slopes of both curves are exactly equal to −0.5, which is the theoretical prediction for a quantum-noise-limited imaging system. The key lesson to be learned from Fig. 6 is that image noise can be controlled by adjusting X-ray tube output. Increasing the output reduces noise; reducing the output increases noise.

It is possible to image a given patient at a range of X-ray tube voltages and adjust the X-ray tube output values so that the resultant CNR is kept constant [22].

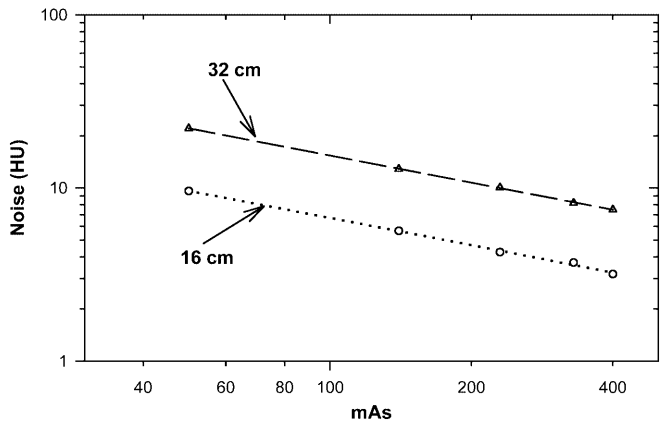

Fig. 6. Noise versus X-ray tube output for a uniform water cylinder. The slopes of these curves are −0.5 which demonstrates that image noise is dominated by quantum mottle

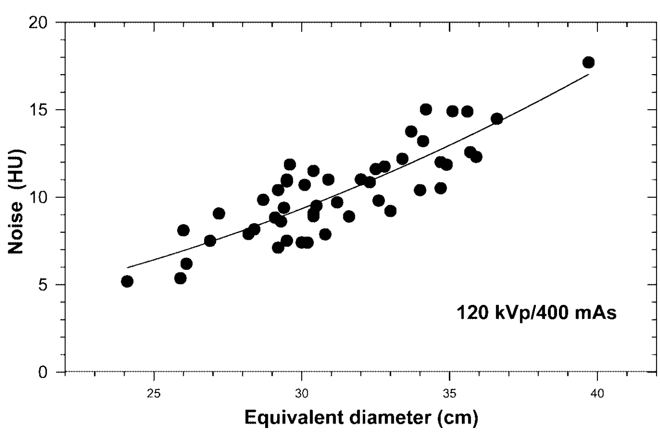

Fig. 5. Measured noise in the liver of patients versus patient size in abdominal CT scanning performed at the same X-ray tube voltage and output

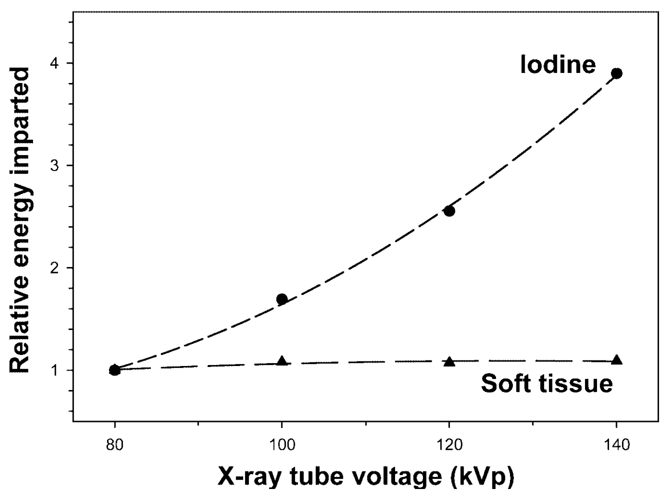

Fig. 7. Energy imparted versus X-ray tube voltage computed at a constant contrast to noise ratio for the specified lesion in a background of water

Under these circumstances, the image quality at each X-ray tube voltage would be identical, but the corresponding patient doses would differ. Figure 7 shows the results obtained for this type of calculation for head CT scans, where the normalization condition is a constant image quality (i.e., CNR). The data in Fig. 7 show that for both high Z and low Z lesions, the lowest patient dose is obtained at 80 kVp. As expected, dose savings from reducing the X-ray tube voltage are much greater for iodine (high contrast) lesions than for soft tissue lesions, since image contrast increases significantly for iodine with reduced photon energy (Fig. 4).

Conclusions

The choice of X-ray tube voltage and output in CT examinations affects both image quality and patient dose. Selection of these X-ray technique factors should be tailored to both the patient size and to the specific imaging task [23, 24]. By ensuring that the amount of radiation used is no more than that required to achieve a satisfactory diagnosis, patient doses will be kept as low as reasonably achievable (ALARA).

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