Continuing Professional Development / Perfectionnement

Canadian Association of Radiologists Radiation Protection Working Group: Review of Radiation Units and the Use of Computed Tomography Dose Indicators in Canada

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Radiation has played a vital role in health care for over a century now. Since Roentgen’s first radiograph in 1895, the medical imaging community has made tremendous advances with imaging technology, leading to earlier detection of disease, minimally invasive approaches to diagnosis and therapy, and, most importantly, improved patient outcome.

Ionizing radiation can directly damage tissue and is recognized by multiple national and international bodies as a weak carcinogen [1–3]. Direct tissue damage is referred to as the deterministic effect of radiation and will occur after the patient receives a radiation dose that exceeds a threshold, whereas the risk of cancer is referred to as a random or stochastic effect, with the probability of developing cancer increasing with increasing radiation dose to the patient. Both deterministic and stochastic effects are a function of radiation dose, a concept with multiple terminologies that has led to significant confusion in practice. With an increased reliance on medical imaging for patient care, both the short- and long-term risks associated with each imaging procedure must be understood by all practitioners. The Canadian Association of Radiologists (CAR) has formed the Radiation Protection Working Group (RPWG) to provide leadership on dose education in Canada. This group aims to develop standards on diagnostic medical radiation protection to decrease patient risk and provide the CAR membership with information, online dose-calculating tools, and a forum to discuss medical radiation dose in Canada.

In this first essay, 3 important concepts will be discussed: radiation risk models, radiation units and the dose delivered by medical imaging systems, and the estimated radiation dose received by the patient. Because computed tomographic (CT) examinations account for the majority of total radiation received by patients, especially when considering the disproportionately lower examination frequency compared with other medical imaging examinations [4,5], these concepts will be discussed with a specific emphasis on dose from common CT examinations.
Radiation Risk Models

Biologic damage from radiation exposure is classified as either deterministic effects on tissue or stochastic effects on cells. Understanding the models used to predict the effects of radiation are valuable towards understanding the radiation protection principles and dose-optimization strategies. Deterministic effects occur after the absorbed dose exceeds a dose threshold that leads to an observable biologic change, which usually is a result of an inflammatory response by the organ or a reaction to cellular death. The severity of the tissue reaction increases with the magnitude of the radiation dose. At a low radiation dose, cellular death may occur but may not result in an inflammatory response nor result in any impairment in organ function. However, as radiation dose exceeds the dose threshold, observable biologic changes will occur. Increased dose will lead to more severe tissue reactions. Larger doses will result in temporary organ impairment or death [1,6,7]. In radiology, the vast majority of procedures should not result in a deterministic effect, because the radiation doses used in diagnostic imaging are magnitudes lower than the threshold doses required for the onset of the following: transient erythema (threshold of approximately 2 Gy), temporary epilation (threshold of approximately 3 Gy), and main erythema (threshold of approximately 6 Gy). Radiologists who perform CT perfusion studies and long interventional procedures should be aware of these dose thresholds levels and formulate an action plan if patient dose exceeds these levels.

Contrary to deterministic effects, stochastic effects do not occur after a dose threshold nor does the severity of the stochastic effect increase with radiation dose. For instance, a 1-Gy exposure that leads to a solid cancer is no different than a 5-Gy exposure that leads to the same solid cancer. Instead, it is believed that the radiation dose will increase the probability of a stochastic effect that occurs, that is, the long-term potential of cancer or hereditary effects will increase with radiation dose. Three dose-risk models have been developed by the Biological Effects of Ionizing Radiation (BEIR) VII [2] of the National Research Council, the linear no-threshold (LNT) model, linear quadratic model, and the linear model with threshold. The LNT model has been widely adopted by international bodies (International Commission on Radiological Protection, National Council on Radiation Protection, National Research Council, and International Atomic Energy Agency) as the method of estimated incidence of cancer and mortality due to radiation exposure. This model has been extrapolated from the cohort of Japanese atomic bomb survivors in Hiroshima and Nagasaki through the Life Span Study (LSS). The LNT model is reviewed periodically when new literature becomes available from medically exposed persons and nuclear energy workers.

The LNT model is derived from atomic bomb survivors, where the radiation dose to the survivors is higher in both rate and quantity, than the radiation used in medical imaging examinations. At low-exposure levels, statistical limitations make it difficult to evaluate cancer risk in humans (BEIR VII [2, page 7]), which leads to a debate on the validity of the LNT model applied to low-dose medical examinations. This debate has led to the “as low as reasonably achievable” (ALARA) principle for radiation protection. Simply put, because we do not know the full extent of biologic changes caused by radiation exposure, we must use as little radiation as possible when obtaining diagnostic quality images. By proceeding in this fashion, if low-level radiation results in stochastic effects, then these effects will be mitigated in future generations.

Radiation Dose Units in Diagnostic Imaging

A variety of radiation terms and units have been described over the years. In diagnostic imaging, the terms exposure (units: roentgen or coulombs per kilogram), activity (units: curie or becquerel), air KERMA (kinetic energy released per unit mass in air, units: gray), absorbed dose (units: rad or gray), equivalent dose (units: rem or sievert), and effective dose (units: rem or sievert) are commonly found in the literature. Our committee agrees with the current literature that as non-SI units, the roentgen, curie, rad, and rem should no longer be used in Canada.

When comparing the radiation output of equipment, the simplest measurement of radiation delivered is the absorbed dose in air. Absorbed dose is the amount of energy deposited within a unit mass, typically measured in air within an ionization chamber. Absorbed dose has the units of joules per kilogram (J/kg) and is designated by the SI unit: gray (Gy). Medical physicists and equipment manufacturers routinely use absorbed dose measures to easily and quickly determine whether equipment is producing radiation within defined standards. The application of absorbed dose to describe radiation exposure in medicine is only useful if the entire patient is uniformly exposed, such as fetal radiation exposure, or if the skin entrance dose is monitored for deterministic effects, which may occur in long fluoroscopic procedures or CT perfusion studies.

The majority of medical imaging procedures focuses on imaging a specific region of interest (eg, a chest CT), and these procedures do not irradiate the entire body. For this reason, absorbed dose is not used to measure patient exposures because this dose does not account for the type of radiation or the radiosensitivities of the irradiated organs.

When describing the risk associated with radiology examinations, the best single descriptor is the effective dose, described in SI units as the sievert (Sv). The sievert incorporates both the type of radiation (x-ray vs gamma ray vs particulate), which have different radiobiologic effectiveness and organ radiosensitivity. The effective dose allows one to compare the radiation risk among different imaging modalities (eg, CT vs nuclear medicine), different body part (CT chest vs CT head), and to compare a medical imaging examination with natural background radiation. For example, the radiation received by a patient in nuclear medicine is related to the activity of the radioisotope administered, target
organ, and pooling of radioactivity in other organs, whereas the radiation received by a patient in CT is related to the dose-length product (DLP) (defined in next section). Neither of these quantities can be directly compared. Instead, physicists have developed methods (Medical Internal Radiation Dose and Monte Carlo simulations) [8,9] to convert activity and DLP to effective dose. These conversions enable the comparison of radiation risk among imaging modalities. For instance, a scintography ventilation perfusion study has an effective dose of 3 millisieverts (mSv) [6], whereas a chest CT has an effective dose of 5-13 mSv [3,9–11]. Although the radiation from the examinations is different, their risks can be compared by using effective dose calculations; in this case, the CT examination has 2-3 times more risk than the nuclear medicine examination. By using the E, we can estimate the risk of fatal cancer induction for a reference subject as 5 excess fatal cancers per 100,000 persons exposed to 1 mSv. This value will be inaccurate for any individual but reflects the increased risk for a large population [1].

CT Dose Indicators and Their Utility in Dose Monitoring

Calculating the effective dose from an ionizing radiation examination for an individual patient is not a trivial task. Many factors will influence the effective dose received by the patient, including imaging modality, imaging parameters (kVp, mAs, filtration), patient habitus (body mass index, weight distribution over imaged area, and implants), and technique (patient positioning and scan area). Due to these factors, the dose delivered to the patient should be referred to as the estimated effective dose, or E.

In CT, E is calculated by multiplying the DLP by a body-specific conversion factor (the k factor) [11–14]. The DLP is the product of the weighted or volume CT dose index (CTDIw, CTDIvol—the radiation delivered to a phantom over 1 gantry rotation and corrected for table pitch) and the scan length of the examination, with the unit: mGy · cm. It is very important to note the CTDIvol and, thus, the DLP is the dose to a phantom of a specific size and when using a specific beam-hardening filter in the system. For an adult head or body scan, the diameter of the phantom used to calculate the CTDIvol is 16 cm and 32 cm, respectively. The beam-hardening filter is used by the system to correct for beam hardening caused by differences in patient thickness. When the patient has a diameter and density different from the phantom size, the CTDIvol can only provide an estimate of patient dose, which may be either under- or overestimated. The underlying physics of this problem is well beyond the scope of this essay. However, it is important for all practitioners to understand that the DLP is a dose indicator. In addition, the k-factor for a specific examination can change with a better understanding of the deleterious biologic effects of ionizing radiation. Understanding the variability in both of these factors is fundamental to realizing that effective dose calculation is an estimate and must be treated as an estimate. For this reason, including E measurements in radiology reports at this time may be counterproductive due to the inaccuracy of the measurement.

Although E provides the estimated effective dose and DLP is a dose indicator, each metric can play a vital role in CT dose awareness and reduction. The E should be used to compare the estimated radiation dose from different imaging modalities, as demonstrated in the previous example. In addition, E may be used to alleviate patient concerns regarding radiation from CT examinations by providing a comparison of E with the amount of annual natural background radiation received by the public (approximately 2 mSv in Canada).

Practitioners need to be familiar with the dose indicator DLP because it is the factor that can be easily monitored. Consequently, it is the CAR recommendation that practitioners review the DLP for each examination performed and compare them with established diagnostic reference levels (DRLs) (defined below), which can be easily accomplished by reviewing the dose report page that accompanies each CT study. In the event that this page is not available, institutions should contact their vendor to enable this feature.

Diagnostic Reference Levels (DRLs) in Canada

DRLs are used in medical imaging by physicians and medical physicists to compare the dose delivered by their imaging protocols with imaging protocols for the same examinations at different institutions. These reference values are defined as the 75th percentile of the patient dose distribution in the form of the dose indicator DLP, from CT scanners across multiple hospitals for the same examination (eg, head CT). Health Canada’s Safety Code 35 provides a table of representative DRL values for common diagnostic procedures [3]. In CT, practitioners can compare typical DLP values at their institution with the values published in Safety Code 35 and in the literature [3,15–18] and listed in Table 1. Institutional DLP values substantially higher than published DRL values may indicate a suboptimal protocol, which may be lowered through imaging technique changes, implementing new technology, and education. It is important to note that DRLs are not static values and should shift over

<table>
<thead>
<tr>
<th>Table 1</th>
<th>Diagnostic reference levels for typical computed tomography examinationsa</th>
</tr>
</thead>
<tbody>
<tr>
<td>Country, reference no.</td>
<td>Head</td>
</tr>
<tr>
<td>United Kingdom [15]</td>
<td>787</td>
</tr>
<tr>
<td>Europe [18]</td>
<td>1050</td>
</tr>
<tr>
<td>Multinational [16]b</td>
<td>527</td>
</tr>
<tr>
<td>British Columbia [19]</td>
<td>1300</td>
</tr>
<tr>
<td>Quebec [20]</td>
<td>1352</td>
</tr>
</tbody>
</table>

a Expressed as dose-length product; the doses were established from only a few scanners from each nation: Canada (2 scanners), United Kingdom (1 scanner), Greece (1 scanner), India (2 scanners), Poland (1 scanner), and Thailand (2 scanners).

b Includes data from Canada, United Kingdom, Greece, India, Poland, and Thailand.
time. The introduction of new “low-dose” CT technology combined with a periodic review and optimization of current protocols will shift the examination dose distribution to lower doses, thus resulting in a lower DRL for a given examination (and technology) over time.

In the DRL values listed in Table 1, Canadian data are only available from British Columbia and Quebec [16,19,20]. Differences in CT technology (number of slices, scanner generation) and CT imaging protocols (eg, the start and end point of a cardiac scan, coverage of CT angiogram) can contribute to large variations in published DRLs, as shown in Table 1. It, therefore, is the CAR RPWG’s mandate to establish DRLs that reflect CT imaging practices across Canada.

The CAR RPWG

This essay highlighted appropriate dose metrics for radiology practice in Canada and described the utility of the DLP as a dose indicator for CT examinations. Future essays by the CAR RPWG will focus on the application of the dose area product as a dose indicator for radiographic and radioscopy examinations and establishing Canadian DRLs for CT, radiography, and radioscopy examinations.

References