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Computed Tomography / Tomodensitométrie

The Emergence of Ultra-Low−Dose Computed Tomography and the Impending Obsolescence of the Plain Radiograph?

Patrick D. McLaughlin, FFR RCSI*, Hugue A. Ouellette, FRCPC, Luck J. Louis, FRCPC, Paul I. Mallinson, FRCR, Timothy O'Connell, FRCPC, John R. Mayo, FRCPC, Peter L. Munk, FRCPC, Savvas Nicolaou, FRCPC

Department of Radiology, Vancouver General Hospital, Vancouver, British Columbia, Canada

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Until recently, computed tomographic (CT) examinations acquired at a radiation exposure equivalent to corresponding plain radiographs would be of grossly substandard image quality, almost certainly resulting in a failure to adequately visualize many anatomic structures. Over the past decade, successive technical breakthroughs have facilitated diagnostic-quality CTs to be acquired at rapidly declining ionizing radiation exposures. Today, the mean effective dose of a radiographic series of the abdomen at 0.7 mSv, pelvis at 0.6 mSv, thoracic and lumbar spine at 1.0 and 1.5 mSv, respectively [1] appear licentious when compared with exposures achieved in recent low-dose CT trials (Table 1). In an era in which low-dose CT has facilitated a 20% reduction in mortality among smokers [7]. and in which doses continue to substantially fall, we propose that radiologists and clinicians should critically reevaluate the risks and benefits of performing many plain radiographic examinations.

* Address for correspondence: Patrick D. McLaughlin, Department of Radiology, Vancouver General Hospital, 855 W 12th Avenue, Vancouver, British Columbia V5Z 1M9, Canada.

E-mail address: mclaughlin.paddy@gmail.com (P. D. McLaughlin).

Technical Background

In brief summary, there have been 3 key developments in CT dose reduction technology that have facilitated the aforementioned trend. Automated exposure control ensures efficient dose delivery by modulating tube current according to patient width and attenuation profile [8−10]. Fixed tube current settings were commonplace in older-generation CT systems and resulted in wider, more attenuating areas, such as the shoulders receiving the same exposure as narrower less attenuating regions such as the upper lungs. More recently, algorithms that modulate CT voltage according to patient size and CT application have also been implemented with good success [11].

After ensuring efficient dose delivery, the largest challenge to obtaining diagnostically acceptable CT images at exposure levels similar to plain radiographs is the severity of random variation in attenuation values that occur within the normal anatomic structures in these images otherwise known as noise. The magnitude of image noise at low CT exposure is fundamentally related to the image reconstruction process [12]. Iterative reconstruction algorithms use a varyingly complex model of the physical characteristics of the x-ray tube, beam, and the 3-dimensional interaction of the x-ray beam within the patient to reduce noise and are clearly better than more traditional methods of reconstruc-

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Table 1
Reported dose values for recent SECT, DSCT, and DECT studies

Authors/year	Study title	Description	SECT	DSCT	DECT
Kerl at al, 2011 [2]	Dose levels at coronary CTA: a comparison of DECT, DSCT, and 16-slice CT	68 patients in each of 3 groups: 16-slice MDCT, DSCT, and DECT underwent coronary CTA	Mean radiation dose (mSv): 12 ± 3.59 DLP (mGy.cm): 760 ± 153.88	Mean radiation dose (mSv): 9.8 ± 4.77 DLP (mGy.cm): 578.07 ± 282.49	Mean radiation dose (mSv) 4.54 ± 1.87 DLP (mGy.cm): 270.81 ± 109.21
Leschka et al, 2008 [3]	Low kilovoltage cardiac DSCT: attenuation, noise, and radiation dose	Dual-source CTCA with retrospective ECG gating in 40 patients at 120 kV per 330 mAs; 20 at 100 kV per 330 mAs; and 20 at 100 kV per 220 mAs.		Estimated ED (mSv): (1) 8.9 ± 1.2 (120 kV per 330 mAs); (2) 2.6.7 ± 0.8 (100 kV per 330 mAs); (3) 3.4.4 ± 0.6 (100 kV per 220 mAs). DLP (mGy.cm): (1) 522 ± 69 (120 kV per 330 mAs); (2) 391 ± 46 (100 kV per 330 mAs); (3) 261 ± 34 (100 kV per 220 mAs).	
De Zordo et al, 2012 [4]	Comparison of image quality and radiation dose of different pulmonary CTA protocols on a 128-slice CT: high-pitch DSCT, DECT, and conventional SECT	Pulmonary CTA performed with 5 protocols: high-pitch DSCT (100 kV and 120kV); DECT (100/140 kV); SECT (100 kV and 120 kV)	Mean radiation dose (mSv): 5.81 (120kV), 3.58 (100kV)	Mean radiation dose (mSv): 2.52 (100kV)	Mean radiation dose (mSv): 4.2
de Broucker et al, 2012 [5]	Single- and dual-source chest CT protocols: levels of radiation dose in routine clinical practice	634 adult outpatient and inpatients undergoing thoracic CT examination with and without a contrast agent	Average DLP (mGy.cm): 211.1	Average DLP (mGy.cm): 97.12	
Ho et al, 2009 [6]	Dual-energy vs single-energy MDCT: measurement of radiation dose using adult abdominal imaging protocols	Radiation dose of dual-energy and single-energy MDCT imaging using adult liver, renal, and aortic imaging protocols	ED (mSv): 9.4-13.8 DLP per phase (mGy.cm): 290.9-614.6		ED (mSv): 22.5-36.4 DLP per phase (mGy.cm): 276.9-997.1

CT = computed tomography; CTA = CT angiography; DECT = dual-energy CT; DLP = dose-length product; DSCT = dual-source CT; ED = effective dose; MDCT = multidetector CT; SD = standard deviation; SECT = single-energy CT; ECG = electrocardiogram.

tion particularly at low CT exposures [13]. Iterative reconstruction algorithms have been pivotal in preserving diagnostic quality at increasingly low radiation exposures, with reductions in the order of 80% reported in the literature, with little or no loss of image quality [14,15]. If a reduction in image quality is acceptable for certain clinical indications significantly more aggressive reductions in CT dose are theoretically possible, to a point, when using iterative reconstruction.

At this point, ultra-low photon flux across the CT detector means that electronic noise contributes substantially more to the reconstructed image. Electronic noise is related to the configuration of detector elements and signal processing pathways at the circuit-board level. Conventional solid-state detectors consisted of a scintillator layer that converts the incoming x-ray photons into visible light and a photodiode array that converts the visible light into an electric current. An analog-to-digital converter is required to digitize the emitted electric current, and, in conventional solid-state or second-generation detectors, this is positioned separately from the scintillator and photodiode array, and typically resides on its own discrete electronic circuit board. Recently, third-generation CT detectors, in which the analog-to-digital converter and photodiode layers are combined have been introduced into clinical practice. These third-generation CT detectors have been described as having an integrated circuit rather than discrete circuit design but have also been termed application-specific integrated circuit and fully