Strategies for Radiation Dose Optimization

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Published online: 30 January 2013
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Abstract This article reviews strategies for radiation dose optimization for computed tomography (CT). A brief overview of dose metrics including computed tomographic dose index, dose length product and effective dose is provided. The impact of age and gender on the sensitivity to radiation is discussed with the aim of tailoring CT acquisition parameters to patient demographics. Dose reduction technologies are reviewed including: tube current modulation, kVp modulation, scan length modification, dynamic z-axis collimation, iterative reconstruction and dual energy. Optimal selection of CT acquisition parameters requires review of clinical information and patient demographics prior to imaging. The authors conclude that with the appropriate application of dose reduction technologies there can be substantial reduction in patient radiation dose while maintaining high diagnostic quality images.

Keywords CT · Radiation dose · Effective dose · Radiation dose reduction techniques · CT pulmonary angiography · CT acquisition protocols

Introduction

The development of computed tomography (CT) in the 1970s and its evolution to the current multi-detector row (MDCT) dual energy capable technology has revolutionized

the role of imaging in medical diagnosis. State of the art CT scanners are able to provide diagnostic, artifact free, whole body scans from head to toe in less than 5 s. These scans provide in vivo images that have similar information content to visual inspection at gross anatomic dissection. The scans can be quickly obtained with minimal patient preparation in essentially all body types and clinical situations. There are minimal restrictions imposed by implanted medical devices or medical monitoring equipment. Finally CT is well accepted by patients and referring clinicians since examinations require minimal patient co-operation, have minimal discomfort and are usually highly accurate.

Although both radiography and CT both utilize ionizing radiation, they have substantially different information content due to differences in the image generation process. Radiography is a measure of differential absorption of the X-ray beam by the patient's anatomy, projected in 2D. In contrast CT obtains 800–1,400 measures of linear X-ray attenuation and combines each projection using a reconstruction algorithm, generating a spatially resolved map of point attenuation values. The map of linear attenuation coefficients is normalized to water, thus providing the CT number. This difference gives CT images a cross sectional viewing perspective that eliminates overlapping shadows that substantially impair the interpretation of the plain radiograph. Thus, CT is sensitive and specific for any disease process that creates macroscopic abnormalities in X-ray attenuation. Additions to basic CT can be used if anatomic abnormalities are not associated with X-ray attenuation differences. An example of this situation is acute pulmonary embolism (PE) where intravascular clot and surrounding flowing blood have similar X-ray attenuation values. Additional contrast can be generated within flowing blood by the injection of iodine containing intravenous (IV) contrast media. This drug transiently elevates the X-ray attenuation

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of flowing blood to allow visualization of intravascular clot. An example of contrast enhanced CT imaging in the chest imaging is CT pulmonary angiography or CTPA.

We can use the CTPA examination to further illustrate the utility of CT imaging for general thoracic CT. The clinical signs and symptoms of acute PE are not specific. Therefore, acute PE is only diagnosed in a minority of patients with clinically suspected PE. The utility of CTPA in suspected acute PE is greatly enhanced as it can diagnose other conditions that may mimic acute PE (e.g. cardiac failure, pneumothorax, pneumonia). As a result, CTPA represents a “one stop shop” for Adult with acute chest pain—suspected PE. Reflecting this, CTPA has been assigned the highest rank, nine in the American College of Radiology (ACR) appropriateness criteria [1] and is considered the first choice imaging examination in these patients with suspected acute PE [2, 3].

Because of the 800–1400 X-ray exposures required in the CT image process, patient radiation dose is 100–400 times greater in chest CT compared to the single view plain chest radiograph. Given the clinical utility and wide accessibility of CT, the high examination radiation dose translates into a large population radiation dose. It has been estimated that CT represents almost 50 % of total medical X-ray radiation dose [4, 5•]. Medical imaging guidelines [1, 6] acknowledge the relatively high radiation dose of the CT examination. Increased awareness of the high level of medical radiation exposure to the population has motivated researchers and equipment manufacturers to evaluate all aspects of the CT imaging chain searching for radiation reduction strategies while retaining the high diagnostic accuracy of the modality [4, 7, 8]. This review provides a brief background on methods to measure CT radiation dose, describes CT radiation dose reduction techniques, discusses tools to enhance CT imaging appropriateness and outlines acquisition protocols that are tailored to subject radiation sensitivity. The CTPA examination, a widely used CT acquisition protocol will be used to illustrate radiation dose optimization techniques in this paper.

Radiation Effect

Biological risk from radiation exposure is generally classified into two macroscopic categories: deterministic and stochastic effects. Deterministic effects occur after the absorbed dose to an organ exceeds a threshold resulting in loss or compromised organ functionality [9, 10]. Deterministic effects require radiation exposures that are approximately two orders of magnitude (>2 Gy) above those received from diagnostic CT (typically <50 mGy). Barring repetitive scanning of the same region that can occur with perfusion studies or inappropriate repetition of diagnostic

scans, such dose levels cannot be delivered in diagnostic CT. Stochastic effects refer the probability of potential long-term cancer or hereditary effects that may occur due to radiation exposure. The life span study (LSS) of nuclear bomb exposed subjects has led to the current linear no-threshold model used to predict stochastic risk to populations exposed to radiation [8, 11]. Two important characteristics of this model are: (1) the severity of biological effect is independent of dose and (2) no “safe” dose threshold exists. Based on the effective dose (E) measure, the excess stochastic risk of fatal cancer induction for a reference subject (70 kg hermaphrodite, age 30 years) is approximately five excess fatalities $100,000^{-1}$ population mSv^{-1} . It is acknowledged that this value is inaccurate for any individual but may be valid for large population [11]. A limitation of risk prediction using E is the use of risk to a reference subject that omits the important subject risk factors of gender and age at exposure to the calculation of risk. The lifetime attributable risk (LAR) metric takes these factors into account and is gaining traction as more appropriate metric for dose tracking and CT scan dose optimization [8, 12–14].

It is noted that despite 100 years of research, the magnitude of the stochastic cancer risk attributable to low dose X-ray radiation exposure remains controversial [15, 16]. This speaks to the weakness of the stochastic cancer inducing effect. However, with the current limited scientific knowledge on the true level of risk, to mitigate potential stochastic risk the ALARA concept (as low as reasonably achievable) has been promoted to guide X-ray exposure levels in medical imaging. Thus a reduction in CT scan radiation dose will decrease the E per scan and lead to an overall reduction in population radiation dose. Using BIER seven tables, it can be demonstrated that the diagnostic benefit of a CTPA examination outweighs the stochastic radiation risk. It is also noted that the benefit to risk ratio is influenced by both age and gender [13•].

Radiation Dose Measurement and Radiation Risk Assessment in CT

The X-ray radiation used in a CT examination can be described using two different metrics. The first metric describes the X-ray energy delivered to a body equivalent polymethyl methacrylate (PMMA) phantom by the scanner per gantry rotation. This is known as the computed tomographic dose index (CTDI) (units: mGy). Although simple and reproducible, CTDI is not directly associated with patient radiation risk. The second metric— E (units: mSv)—incorporates the radiosensitivity of the object scanned, thus providing a basic measure of risk.

Computed tomographic dose index is an easily obtainable metric on radiation dose output to confirm CT

equipment is operating according to manufacturer specifications. It is routinely measured by Medical Physicists at regular intervals to confirm consistent system performance [17, 18]. For step and shoot acquisition, this metric is calculated by measuring the absorbed dose (J kg^{-1} , units: mGy) at the center and periphery of a standardized PMMA phantom from a single slice [19]. This simple measure incorporates the dose distribution from the primary beam and secondary scattered radiation. For step and shoot scans the CTDI is referred to as CTDI weighted (CTDI_w, units: mGy). For helical acquisitions the influence of the helical pitch must be considered and this is reflected in a second metric, CTDI volumetric (CTDI_{vol}, units: mGy) [20, 21].

It is noted that the CTDI it is not the true patient dose, but the dose delivered to a standardized phantom [22]. By multiplying the CTDI_{w/voI} by the scan length in centimeters, the applied radiation dose to a scanned volume is calculated. This is known as the dose length product (DLP). Thus the DLP extends the CTDI_{w/voI} from a single rotation to the entire scanned volume.

Recently, AAPM Task Group 111 published a report discussing the future of CT dosimetry [23], a defined new metrics for radiation evaluation of helical and cone-beam CT. At the present time, these parameters are beyond clinical discussion, and are mentioned here solely for completeness.

Effective dose is the only dose metric that incorporates stochastic risk and allows for comparison of risk between medical imaging examinations and with other sources of radiation exposure (nuclear medicine, radiation therapy, natural background, air travel). Mathematically, E is simply calculated by determining the average energy deposited within each organ, multiplied by the organ's radiosensitivity factor and then summed over all organs [11]. In practice, a multitude of factors will influence absorbed organ dose making this calculation difficult and replete with uncertainty and assumptions. A description of each factor is beyond the scope of this review; however, a short list of major factors is provided in Table 1.

Monte Carlo software has been developed to estimate patient specific organ dose, and thus provide estimates of a

patient's E [24, 25]. Accurate calculation of E using patient specific Monte Carlo methods require accurate segmentation of individual patient organs and the use of computationally intensive algorithms. Because of these two necessities, widespread application of Monte Carlo software is currently impractical and only appropriate to the research setting. A simplified approach has been developed for the clinical setting using a generic anatomically specific conversion factor. Using this approach, E can be estimated by multiplying the DLP by a body specific conversion factor (the “ k factor”) [26–29]. Body region specific k factors for head and neck, chest, abdomen and pelvis, trunk and extremities have been determined using Monte Carlo simulation in reference subjects. These body region specific k factors allow conversion of DLP values to E . There is a range of k factor values depending on the assumptions made regarding the scanned volume of reference subject (e.g. cardiac k factor versus chest k factor). This highlights the uncertainty inherent with this approach and why these factors should not be used to calculate “patient specific dose.” In addition to uncertainty in the k factor, it is known the CTDI is patient size dependent that is not accounted for in the DLP [5•, 30]. Finally, the organ specific radiosensitivity factors used to calculate E change over time as our understanding of radiosensitivity increases. A good example is the change for gonad weighting factor from 0.2 to 0.08 from ICRP60 to ICRP103 [11, 31]. The inaccuracy of current conversion estimates of E makes it inadvisable to include these measures in radiology reports. However, the use of DLP as an exposure metric may be useful to increase awareness of radiation dose issues and feedback for the optimization of scan protocols, via establishment and comparison to published diagnostic reference levels (DRLs).

Radiation Sensitivity of Patients

Patients differ in their sensitivity to radiation based on their age and gender. The most susceptible are children who

Table 1 List of factors affecting organ dose

Patient related	Dose reduction technologies	Protocol attributes	Scanner physical attributes
Organ motion (lung, cardiac, digestive)	Tube current modulation (Z, angular)	Beam collimation (scan volume)	Variation of tube potential during exposure
Organ position and size	Peripheral dose reduction	kVp, mA, rotation time, pitch	Tube inherent filtration
Patient habitus in supine position	Dynamic Z-collimation (shutter action)	Helical, axial, cone	Beam quality (pre-filtration)
Implants		Noise index/mAs reference	
Sex, age		Contrast medium location and density	

have a greater percentage of replicating tissue and a longer life expectancy to manifest the stochastic carcinogenic risk [32]. Risk decreases with increasing age, being very low above the age of 80 years. Therefore, the risk of radiation is greatly attenuated by increasing age and radiation dose reduction strategies that might impair image quality are more rational in younger patients and may be inadvisable in older individuals. Up to the age of 60 years, females are more sensitive to radiation than males of the same age. Although this effect is partially modulated by breast tissue in females up to the age of 50 years, above this age breast tissue is relatively radiation insensitive and the chest organs at most risk in both males and females are bone marrow and lung parenchyma [33].

Radiation Dose Reduction Strategies

Tube Current Modulation

The thoracic region of the body in most patients can be visualized as a series of elliptical volumes, with an incrementally changing effective diameter. Generally speaking, the short axis (anterior–posterior) is shorter than the left–right axis of the patient; in addition, average tissue density within the beam changes with each projection (i.e. the long axis through the shoulders is of higher average density than a long-axis path through the apex of both lungs). In highly attenuating projections insufficient photons reach the detectors and result in very noisy image reconstruction. The reconstruction algorithm magnifies the noise in these projections, resulting in streaks in the image [34]. Therefore, optimal image quality is achieved at lowest dose when all projections have comparable numbers of detected X-ray photons. This can be achieved by varying the X-ray tube current dynamically throughout the CT scan in the x, y and z dimensions. This feature, known as tube current modulation, produces chest CT images with more uniform image noise. Angular modulation (x, y) has been shown to reduce chest dose by 22 %, while a 26 % dose reduction was reported with z-axis modulation [35, 36].

Manufacturers have implemented this feature using two approaches. Some measure the relative attenuation of the patient by the use of two tightly collimated pre-scans (referred here as digital radiographs for the short and long axis). The z-axis modulation is planned from the frontal view, with the x and y axes planned from a combination of the frontal and lateral views. The second approach uses a frontal pre-scan planning digital radiograph only. The z-axis modulation is planned off this pre scan digital radiograph while the x and y axes modulation is determined by real time monitoring of the attenuation of the previous X-ray tube rotation. It is noted that bismuth breast

shields placed over a female patient's breasts between the pre scan digital radiograph and the scan acquisition will not interfere with the first tube current modulation technique described using the frontal and lateral approach. However, bismuth breast shields will interfere with the real-time x and y axes modulation described in the second tube current modulation technique. As such, bismuth shields should not be used with CT systems employing real-time tube current modulations. Generally, use of bismuth shields is discouraged when tube current modulation and peripheral dose reduction techniques are available [37–39].

kVp Modulation

As a generic rule of thumb, the X-ray tube output change is related to the square of the ratio between the new kVp and the reference kVp. Thus, at constant tube current (mA) and tube rotation (s), decreasing kVp from 120 to 80 decreases radiation dose in air from 58.8 to 21.9 mGy using a GE VCT64 [technique: 1 s, 260 mA, body filter, 8×5 mm, (GE Medical Systems, Milwaukee, Wisconsin)]. In the situation of contrast enhanced CT, iodine containing contrast media is administered to improve contrast sensitivity to abnormal vasculature or clot within blood vessels. Iodine within the contrast media has a k-absorption edge of 33.2 keV. By matching the average energy of the X-ray tube spectrum to the k-edge X-ray absorption, radiation dose can be reduced while image contrast is increased. Increasing photon energy substantially beyond this value increases scatter radiation and dose and decreases image contrast. As a general rule the average X-ray beam energy is 1/3–1/2 the tube potential, depending on beam filtration [40]. Thus, an 80 kVp beam has an average energy range of 24–40 keV, while a 120 kVp beam has an average energy range of 30–60 keV. Thus at 80 kVp, the k-edge for iodine is matched and less scatter radiation occurs. The drawback of this technique is fewer photons reaching the detector due to reduced beam penetration. For this reason, low kVp is inappropriate in large patients as there is near complete attenuation of the photon beam, leading to noisy non-diagnostic images. Thus, lower tube voltage can be used in small patients while preserving image quality, but will yield excessively noisy images in larger patients. Small patients will receive substantial radiation dose reduction with this approach and image quality improvement will be greatest in contrast enhanced CT scans. This has lead to BMI and weight-based protocols to optimize delivered dose.

Scan Length Adjustment

Since the DLP is linearly related to the scan length, limiting the scan length reduces irradiated patient volume and

thus radiation dose. Using the example of the CTPA examination, pulmonary arteries of interest in CTPA are central, segmental and sub segmental in size, vessels that are found from the level of the aortic arch to the level of the lowest hemi-diaphragm in majority of patients. Interestingly, this restricted scan volume for CTPA examinations was routinely used in the 1990s for single slice scanners due to slow scan speeds that required 20 s breath-holds. Initially, with the increased X-ray power of MDCT systems the scan length was increased to include the total thoracic volume from the lung apex to below the posterior costophrenic angle. However, this approach substantially increased radiation dose due to the increased scanned volume. In the interest of dose reduction, a decrease in scan length of MDCT CTPA scans has been suggested. It has been found that scan length adjustment from just above the aortic arch to just below the heart will maintain 98 % diagnostic accuracy with a dose reduction of 37 % [41, 42]. A potential limitation of this technique is reduced accuracy for alternate diagnosis as the entire chest is not scanned. For this reason, this technique is best employed in young or pregnant subjects with normal chest radiographs and minimal suspicion of a diagnosis other than PE [43]. In other CT scan protocols, a similar approach to limitation of the scan length to cover only the region of diagnostic interest will produce similar reductions in radiation dose.

Z-Axis Overscan

In helical scanning, the first image can only be reconstructed at the first point in the helix where 180 plus fan angle projections are available. Therefore, wasted X-ray radiation is delivered at the beginning and the end of the scan. The proportion of wasted radiation is greatest in small volume acquisitions. This occurs in small patients or when large volumes are acquired as a group of smaller volumes [44, 45]. This effect is known as z-axis overscan and can be eliminated by selectively moving the collimator blades in the z-axis direction at the beginning and end of the scanned volume. This technique commonly referred to as a collimator shutter action is available on all new scanners.

Iterative Reconstruction

For the last 35 years the most commonly used CT reconstruction algorithm has been filtered back projection (FBP). The susceptibility of FBP to image noise relative to iterative reconstruction (IR) techniques has been well known for decades; however, the computational simplicity of the FBP reconstruction allowed sub-second image reconstruction which facilitate rapid clinical throughput. Iterative reconstruction techniques are computationally more

complex and subsecond image reconstruction was not possible with the computer power available in the past. Improvements in computer technology have permitted the introduction of this reconstruction technique on MDCT scanners in the last 5 years.

Various modifications of IR are being developed and refined by different CT manufacturers. IR algorithms can be performed on the image data, on the raw projection data from the scanner, or both. Iterative reconstruction in image space (IRIS) (Siemens Healthcare, Erlangen, Germany) is an example of an IR algorithm which uses the image data alone. Iterative reconstruction algorithms which use both the image and raw data include adaptive statistical iterative reconstruction (ASIR) (General Electric Healthcare, Milwaukee, Wisconsin), Sinogram Affirmed Iterative Reconstruction (SAFIRE) (Siemens Healthcare, Erlangen, Germany), Adaptive Iterative Dose Reduction 3D (AIDR-3D) (Toshiba Medical Systems, Tustin, California), and iDose (Phillips Healthcare, Best, the Netherlands). Many of these algorithms including ASIR blend traditional FBP data with IR data and are, therefore, referred to as hybrid algorithms. Radiation dose reduction of at least 25 % can be achieved with IR techniques and images typically have preserved image quality and good low-contrast detail (Fig. 1) [46].

Pilot studies with ASIR [47, 48], with IRIS [49] and with SAFIRE [50] found that radiation dose can successfully be reduced by 50 % or greater in abdominal CT without significantly affecting image quality. However, the use of high IR strengths in low dose CT images often results in a subjectively different image quality and noise texture to images reconstructed with FBP. The different qualities of IR images have been described as being “waxy” or “plastic” in the case of early IR algorithms to mildly “mottled” or “pixelated” in the case of later generations of IR. Expert opinion in this area suggests that imagers tend to adapt to the new quality of these images in a relatively short period of time [51].

A disadvantage of hybrid IR algorithms is that they are reliant to some degree on image data. The next generation of IR algorithms operate using the raw projection data only. This newer generation of IR algorithms are referred to as “pure” IR algorithms. Model-based iterative reconstruction (MBIR) is a commercially available “pure” IR technique developed by GE Healthcare. Model-based iterative reconstruction incorporates a physical model of the CT system into the reconstruction process including the characteristics of the focal spot, the X-ray fan beam, the 3-D interaction of the X-ray beam within the patient and the 2-D interaction of the X-ray beam within the detector [51, 52].

Pure IR algorithms have not been practical for commercial CT scanners until recently due to constraints in computing power and reconstruction technology.

Fig. 1 58 year old female patient with acute right lower lobe subsegmental pulmonary emboli (**a, c**). Transverse CT images reconstructed with FBP showing moderate noise in the pulmonary artery but satisfactory visualization of the subsegmental pulmonary embolism (*arrow*) (**b, d**). Transverse CT images reconstructed with SAFIRE (strength = 2) showing reduced noise in the pulmonary artery and improved conspicuity of the subsegmental pulmonary embolism (*arrow*)

Computing demands for MBIR remain significant and reconstruction time for MBIR images in today's commercially available CT systems is still in the order of 3–4 datasets per hour despite the use of parallel processing technology [52]. Pilot studies examining the benefits of pure IR algorithms with the aim of introducing it into clinical practice are now in progress. Emerging data suggests that low dose abdominal CT using pure IR algorithms outperforms hybrid reconstruction algorithms such as ASIR and also outperforms FBP in both subjective image quality indices and objective image noise scores facilitating dose reductions of approximately 80 % [53••].

Dual Energy

Dual Energy CT (DECT) allows material decomposition of soft tissue, iodine and air within the chest. Thirty years ago, the clinical utility of this technique was recognized. However, at that early stage of CT scanner development, adequate image quality could not be attained since existing CT X-ray tubes could not generate sufficient current for acceptable photon flux and image quality for the low kVp acquisition (e.g. 80 kVp). Recent developments in CT X-ray tube technology have enabled this application on current scanners. This technique provides a new contrast mechanism on CT, material decomposition. The initial targets have been uric acid crystal identification for the non-invasive diagnosis of gout (Fig. 2) or characterization of renal stones and iodine content measurement for the assessment of tissue perfusion (Fig. 3). The characteristic

difference in attenuation of the numerous types of renal calculi with dual energy imaging allows more accurate determination of stone composition which potentially facilitates more appropriate management in patients with uric acid (UA) containing calculi who may benefit from medical management versus those with cystine and certain calcium stones which may be more resistant to shock wave lithotripsy [52]. In CTPA scans, the quantification of lung parenchymal perfusion via lung tissue iodine concentration should assist in the diagnosis of PE.

Currently, there are two approaches to DECT mandated by the X-ray tube configuration of the scanner, dual source (DS-DECT) or single source (SS-DECT). Both configurations have associated advantages and limitations which are beyond the scope of this review. DS-DECT systems have two independent X-ray tubes, which operate at different tube potentials (i.e. tube A—80 kVp, tube B—140 kVp), while single source systems employ rapid switching between kVps. Using the example of CTPA, both techniques have demonstrated acceptable image quality [54–56]. In CTPA studies there are two potential applications of dual energy techniques, radiation dose reduction with improved diagnostic accuracy and reduction in contrast media volume with preserved diagnostic accuracy. Regarding radiation dose reduction, second generation DS-DECT has shown a 28 % reduction in patient dose compared with single source 120 kVp protocols with improved image noise, vessel contrast and diagnostic confidence [56, 57]. In a randomized clinical trial Yuan et al. [54] demonstrated that DSCT reconstructed at a 50 keV energy allowed significant

Fig. 2 47 year old male patient with gout. **a** 3D Dual energy CT image demonstrating extensive monosodium urate crystal deposits (*green*) in the metatarsal phalangeal joints, intertarsal joints and Achilles tendon bilaterally. **b** Follow up 3D dual energy CT image acquired 12 months later demonstrating a marked reduction in the monosodium urate deposits (*green*) after successful diet and medical therapy (Color figure online)

Fig. 3 44 year old male patient with chronic pulmonary emboli. **a** Transverse CT image demonstrating a large clot burden in the right lower and middle lobe pulmonary arteries (*arrows*). **b** Transverse dual energy CT image with iodine mapping (*red*) demonstrating a corresponding reduction in iodine volume in the right middle and right lower lobes (*arrows*) (Color figure online)

reduction in iodine load at CTPA while maintaining compatible signal to noise ratio, contrast to noise ratio and similar effective radiation dose. Further research should increase the impact of DECT on all CT examinations.

Clinical Appropriateness

Patient requisition should be reviewed by radiologist prior to booking to determine patient age and determine the level of radiation dose reduction used:

1. Highest radiation dose reduction; < 30 years of age.
120–100–80 kVp depending on patients BMI or automatic tube voltage selection.
150 mA tube current with x, y, z tube current modulation.
The widest section thickness appropriate to the clinical question should be employed, IR at minimum 40 % or

level 2 strength, higher strength as clinical familiarity with IR images improves.

2. 120–100–80 kVp depending on patients BMI or automatic tube voltage selection.
200 mA tube current with x, y, z tube current modulation.
The widest section thickness appropriate to the clinical question should be employed, IR at 40 % or level 2 strength
3. Lowest radiation reduction; >60 years of age.
120 kVp
200 mA tube current with x, y, z tube current modulation.
Section thickness appropriate to the clinical question, IR or FBP reconstruction depending on the radiologist's preference.

When validated clinical prediction rules are available, these should be employed to improve patient selection for

imaging studies. An example of such rules for suspected acute PE are the Pulmonary Embolism Rule-Out criteria, Wells score and Geneva score. Laboratory tests may also be used to improve patient selection for CT imaging studies (e.g. D-dimer level testing of the blood for suspected acute PE studies) [58].

Summary

Radiation dose is an important issue for CT studies as they are commonly performed in patients of all age groups. Justification of all CT examinations is critical and is monitored by review of all clinical requisitions prior to scanning. Education of referring clinicians on the utility of clinical prediction tools and predictive laboratory test values will improve CT examination justification and ensure that CT is employed for maximum benefit. Audit of the positive rate of CT examinations is also useful to validate and improve justification. Radiation dose reduction techniques are most important in young patients due to their increased radiation susceptibility. Tube current modulation should be used in all cases with tube voltage modulation in younger and physically smaller individuals. Tube voltage modulation may improve image quality in contrast enhanced CT examinations due to improved matching of the mean energy of the X-ray beam to the k-edge of iodine. Iterative reconstruction with associated radiation dose adjustment is recommended for younger patients but may be used in all patients once familiarity has been achieved with the new look of the images. Further research is needed on the utility of the various strengths of IR to ensure that this new reconstruction technology is appropriately employed. Patients requiring contrast enhanced CT protocols with impaired renal function may benefit from dual energy acquisitions with reduction in iodine load by approximately 50 %. Radiologists should be familiar with the DLP metric reported on the scanner console and should monitor this metric in their clinical practice. Ideally, this metric should decline over the next few years as protocols incorporating patient BMI and age, and utilizing all available dose reduction technologies becomes the standard of practice for all CT examinations.

Disclosure Y Thakur: none; PD McLaughlin: none; JR Mayo: receives honoraria from Siemens Medical Solutions for speaking at Siemens sponsored meetings and receives funding for equipment evaluation from Siemens Medical Solutions.

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