
Theoretical Background

Chapter two: Theoretical Background

2.1. CT principles

Medical imaging provides tremendous and undeniable benefits for patients in modern health care. Imaging is used for a broad range of tasks, including disease detection, classification, prognostic staging, treatment planning, and validation of therapeutic responses. During recent years, substantial developments have been made in imaging techniques with progress continuing today.

Computed tomography (CT) (Hounsfield, 1976) and magnetic resonance imaging (MRI) (Damadian, 1971) have become important imaging techniques since their introduction in clinical practice in the 1970s and 1980s, respectively. They have supplemented, and in some cases even replaced, planar X-ray, nuclear medicine, and ultrasound (Mattsson et al., 2010).

After the introduction of multi-slice CT (MSCT) in 1997 (Hu, 1999), the number of slices acquired per rotation has rapidly increased from 4 up to 8, 16, 32, 40, 64, 128, and 320 (Hsieh, 2009). The primary advantage of MSCT is improved temporal (<250 ms) and spatial resolution (<0.5 mm) and shorter scan times (Flohr and Ohnesorge, 2007). However, shortcomings and pitfalls with increased detector width have revealed more scattered radiation, introduced cone-beam artifacts, and helical over-scanning. Today, beam widths are up to 160 mm, making the current paradigm for characterizing radiation absorbed dose in CT by means of

the computed tomography dose index (CTDI) no longer appropriate (American Association of Physicists in Medicine: AAPM, 2010).

The objectives in CT development have changed from increasing the number of slices to focusing on improvements in X-ray tube performance, detector efficiency, and data processing (Fleischmann and Boas, 2011). Since 2006, a new scanner technology using two X-ray sources and two detectors simultaneously, dual source CT, has been available (Johnson and Kalender, 2011). The technology has provided further improvements in scan speed and temporal resolution (0.28 s rotation time and 75 ms temporal resolution). By utilizing dual energy CT (DECT), using either dual source or kV-switching, and advanced post-processing and visualization, new clinical applications have been found.

The advantage with DECT is that the properties of X-ray attenuation change at different energies, which are used to differentiate materials, including iodine, calcium, and uric acid crystals. In recent years, iterative reconstruction methods have been introduced that provide great potential for improving image quality and reduced radiation doses (Hsieh, 2009).

Large-area flat panel detectors in combination with cone beam X-ray fields (cone beam CT, CBCT) are now used more frequently as an alternative to conventional CT (Gupta et al., 2008). Examples of CBCT applications include interventional

and intra-operative imaging, image-guided external tumor therapy, maxillofacial scanning, and breast imaging. Moreover, flat panel detector technology is also applied to standard CT.

As mentioned above, the coverage of a large volume per rotation has led to demands for a new framework for characterizing the absorbed dose.

In addition to CT development, diagnostic imaging has evolved from standalone techniques to combined (hybrid) imaging (Beekman and Hutton, 2007; Townsend, 2008). The nuclear medicine methods single photon emission computed tomography (SPECT) (Kuhl and Edwards, 1963) and positron emission tomography (PET) (Brownell et al., 1971) have been shown to improve diagnostic accuracy in a variety of clinical applications when used in combination with CT (Hicks et al., 2007; Townsend, 2008). The complementary anatomical, functional, and molecular information provided by these hybrid techniques has proven clinical importance, e.g., The molecular process of a tumor can be accurately identified and localized to a specific tissue or organ.

The improvements in CT technology have led to an increased use of CT, and it has replaced several radiographic examinations. A report from the Swedish Radiation Safety Authority (Almen et al., 2008) showed that the number of CT investigations in Sweden increased by 100% between 1995 and 2005. The report also showed that CT and nuclear

medicine constituted 16% of all radiological investigations (excluding mammography and dental examinations) and contributed to 64% of the collective effective dose in Sweden in 2005.

The National Council on Radiation Protection and Measurements (NCRP, 2009) in the USA reported that CT and nuclear medicine constituted 22% of all radiological investigations but 75% of the collective US radiation effective dose in 2006.

The significant increase in the use of CT, alone or combined with SPECT or PET, has raised concerns about patient radiation exposure and the consequent increased risk of malignancy later in life (Brenner and Hall, 2007; International Commission on Radiological Protection: ICRP, 2000; 2007b; United Nations Scientific Committee on the Effects of Atomic Radiation: UNSCEAR, 2008).

Another problem is that some of the PET and SPECT investigations give a high patient effective dose compared to the majority of planar X-ray investigations.

A

B

Figure 2.1 Although CT and nuclear medicine examinations contribute a relatively low percentage of the total number of diagnostic radiological examinations (excluding mammography and dental examinations) (A), CT and nuclear medicine contribute a high proportion of the collective effective dose (B). The figure refers to data from Sweden in 2005 (Almen et al., 2008).

The gradually increasing awareness of radiation exposure mainly from CT

Examination has forced manufacturers to develop techniques to reduce radiation doses. The implementation of these methods, as well as recommendations from authorities, requires close collaboration between medical physicists, manufacturers, radiologists, nuclear medicine physicians, technologists, and referring physicians in order to be effective. The challenge is to establish sufficient image quality for a specific diagnostic task with the lowest effective dose to the patient.

2.2 dosimetry:

CT scanners generate cross-sectional images by measuring X-ray attenuation properties from multiple directions around the region of interest when the X-ray tube and detector are rotated around the object (Kalender, 2005). When the X-rays penetrate the object, parts of its energy is absorbed by the object. The amount of energy imparted per

unit mass at a point is expressed in terms of absorbed dose as defined by the International Commission on Radiation Units and Measurement(ICRU, 1998).

The absorbed dose is the fundamental dissymmetric quantity, and its unit is joule per kilogram, denoted as gray (Gy). To assess radiation exposure to humans and correlate it with the risk of exposure, mean absorbed dose in an organ or tissue is used (ICRP, 2007b). Based on the dose quantities prescribed by the ICRU and ICRP, the International Atomic Energy Agency (IAEA) has established an international code of practice for dissymmetry in diagnostic radiology (IAEA, 2007).

SPECT (Zeng et al. 2004) and PET (Lewellen and Karp, 2004) are imaging

Techniques using radiopharmaceuticals. The tracer compounds are labeled with single photon or positron-emitting radionuclide's, respectively, and injected into the subject prior to the investigation. The radionuclide in the radiotracer decays, and the resulting photons are detected by surrounding external detectors. Parts of the emitted energy from the decays will be absorbed by the body, which is expressed in terms of mean absorbed dose. The signal acquired by the detectors is used to reconstruct the radiopharmaceutical distribution throughout the patient.

2.2.1 The Principles of CT Dosimetry :

For CT, estimates of absorbed doses to organs and tissues and effective doses are based on two quantities: CTDI and dose-length product (DLP) (AAPM, 2008). The CTDI concept was originally introduced for single slice axial scanning (Shoppe et al., 1981). CTDI represents the average absorbed dose along the z-axis (table feed direction) from a series of contiguous irradiations. The most commonly used index is CTDI₁₀₀, which refers to absorbed dose in air or in cylindrical polymethyl methacrylate phantoms (15 cm in length) representing head (16 cm in diameter) and body (32 cm in diameter). The International Electrotechnical Commission (IEC, 2009) has defined CTDI₁₀₀ as the absorbed dose integrated over a length of 100 mm for a single axial scan using a pencil ionization chamber with an active length of 100 mm, divided by the collimated beam width (if $n \cdot T < 100$ mm) or 100 mm (if $n \cdot T \geq 100$ mm):

$$CTDI_{100} = \frac{1}{\min(n \cdot T, 100 \text{ mm})} \int_{-50 \text{ mm}}^{+50 \text{ mm}} D(z) dz \quad (2.1)$$

Where n is the number of slices per rotation, T is the nominal slice thickness, and $D(z)$ is the absorbed dose profile along the z -axis. To account for spatial variation of the absorbed dose in the scan plane (x, y), a weighted dose index (CTDI_w) was introduced (Leitz et al., 1995):

$$CTDI_w = \frac{1}{3} CTDI_{100(\text{central})} + \frac{2}{3} CTDI_{100(\text{peripheral})} \quad (2.2)$$

To take axial scan spacing into account, CTDI by volume (CTDI_{vol}) was introduced (Bongartz et al., 2004):

$$CTDI_{vol} = \frac{CTDI}{pitch} \quad (2.3)$$

Where pitch is defined as the ratio of the table transportation per rotation to the collimated beam width (Silverman, 2001). CTDI_{vol} is expressed in mGy and is displayed on the CT consoles. The CTDI_{vol} is a measure of the radiation output of a CT scanner and represents an estimation of the average absorbed dose within the irradiated volume of an object of similar attenuation to the CTDI phantom. CTDI_{vol} needs to be adjusted for patient size because it does not represent the average absorbed dose for objects of substantially different size or shape (AAPM, 2011).

To better represent the overall energy delivered for an entire CT exam, DLP expressed in mGy·cm was introduced (Bongartz et al., 2004):

$$DLP = CTDI_{vol} \cdot L \quad (2.4)$$

Where L is the scan length. DLP is a measure of the total energy deposited in the phantom or patient.

The quantity effective dose is the sum of weighted equivalent doses in the principal tissues and organs of the body (ICRP, 1991; 2007b). The different tissues and organs have been assigned a tissue weighting factor that reflect the radio sensitivity. The equivalent dose expresses the biological impact of a given type of radiation. Consequently,

effective dose reflects the stochastic risk, such as cancer induction, and the unit is sievert (Sv) (ICRP, 1991). Broad estimates of the effective dose can be obtained by multiplying DLP by a conversion factor (k) appropriate to different anatomical regions (Bongatz et al., 2004; Huda et al., 2008; Shrimpton, 2004):

$$E = DLP \cdot k \quad (2.5)$$

The conversion factors (Table 2.1) are averaged over all photon energy distributions used in different scanners, and obtained from Monte Carlo simulation and mathematically describable phantoms representing adult and pediatric patients. The factors are useful for quick dose estimates and for large patient groups. Conversion factors for DLP to effective dose for different tube voltages, regions, and ages based on the latest tissue-weighting factors from ICRP (2007b) were recently determined by

Deak et al. (2010) and are valid for a Siemens Sensation 64 CT scanner (Table 2.1).

The tissue weighting factors were modified by ICRP due to new available scientific data. For a more detailed assessment of effective dose and organ absorbed doses, dose assessment software, such as CT-Expo (Stamm and Nagel, 2002) and Impact CT patient dissymmetry calculator (Keat, 2011), are recommended. The effective dose from a CT investigation typically ranges from 2 mSv (head) to 10

mSv (abdomen and pelvis), but with large variations between patients and hospitals.

A total body investigation (brain, chest, abdomen, and pelvis) provides about 20 mSv (ICRP, 2007a). This is roughly a factor of 10 to 100 higher than typical conventional planar X-ray investigations (range of 0.01-10 mSv). For example, the effective dose for a hand radiograph is less than 0.1 mSv. CT doses are highly dependent on the characteristics of the CT scanner, patient size, anatomical region under investigation, and scanning parameters used in each examination. For some individuals, local organ and tissue doses from a CT investigation can be up to 100 mSv (ICRP, 2007a; UNSCEAR, 2010).

Table 2.1 Conversion factor k (mSv mGy⁻¹ cm⁻¹) for DLP to effective dose for various body regions in adults and pediatrics patients of various ages. Conversion factors based on tissue weighting factors from ICRP Publication 60 (1991) valid for single-slice CT scanners have been published by Bongartz et al. (2004), and conversion factors based on tissue-weighting

Factors from ICRP Publication 103 (2007b) valid for a 64-slice CT scanner have been published by Deak et al. (2010)

ICRP6

0

Region of body	Adult	Adult	10 year olds	5 year olds	1 year	Newborns
Head	0.0023	0.0019	0.0027	0.0035	0.0054	0.0087
Neck	0.0054	0.0052	0.0094	0.0121	0.0168	0.021
Chest	0.019	0.0146	0.0237	0.0323	0.0482	0.0739
Abdomen	0.017	0.0153	0.0249	0.0357	0.053	0.0841
Pelvis	0.017	0.01291	0.0219	0.03	0.0446	0.0701
Legs	0.0008	—	—	—	—	—

The CTDI methodology, which is based on the assumption that the beam width in the z-axis is substantially smaller than 100 mm, currently has some limitations (Boone, 2007). As the radiation beam widths of MSCT scanners get wider and various CBCT systems are introduced, this method of estimation becomes more inappropriate. When the primary beam is nearly the same size or larger than the length of the

probe, the dose will be underestimated because a portion of leakage and scattered radiation will not be measured. A new measurement paradigm was recently proposed (AAPM, 2010; Dixon and Boone, 2010). Implementation of these recommendations is currently being evaluated by the AAPM task group 200 and ICRU. A new polyethylene-based phantom has been designed, measuring 30 cm in diameter and 60 cm in length, to quantify the equilibrium dose. The equilibrium dose can be measured by a small radiation detector in the phantom that is long enough to capture the entire scatter tails.

2.2.2 Methodologies for dose reduction

Recent technological advances have increased the number of clinical applications for CT. Due to the increased number of examinations CT is the largest source of medical radiation exposure to the general population (Almen et al., 2008).

The reduction in CT doses during recent years has been significant, mainly due to improved technology from manufacturers and increasing awareness at the operator level of the importance of acquisition parameters for patient dose (ICRP, 2007a). Technological advancements have resulted in an increased scan speed, a capacity to provide large scan coverage, better contrast utilization, less image noise, increased spatial resolution, and improved temporal resolution. For example, today a cardiac CT scan can be performed at an effective dose of roughly 1 mSv using

electrocardiogram-based tube current modulation (TCM). A decade ago the effective dose for a cardiac examination was up to 20 mSv, depending on scanner-specific factors and acquisition protocol (Mayo and Leipsic, 2009).

Several methods are available to optimize and minimize the radiation absorbed doses in CT (Kalender et al., 2008; Kalra et al., 2004; Mattsson and Soderberg, 2011a).

The *scanning parameters* should be optimized for each specific examination and special efforts should be made with pediatric CT protocols (Frush, 2008).

A number of scanning parameters influence patient radiation dose and image quality: tube current, tube voltage, filtration, collimation, reconstruction method, reconstruction filter, slice thickness, pitch, and scanning length (Kalra et al., 2004). The operator can monitor most of these parameters and modify them to obtain the necessary image quality with a minimal absorbed dose to the patient. A simple relationship exists between the tube load (the product of tube current and exposure time per rotation, mAs) and radiation dose to the patient. A 50% reduction in tube load reduces the radiation dose by half (and reduces the detector signal), but also increases the noise level by a factor of 2.

An adequate mAs level can be determined using dose reduction simulation software (Soderberg et al., 2010). The software adds artificial noise to the CT raw data to simulate a scan acquired with lower dose (mAs). The tube voltage

determines the energy of the emitted photons from the X-ray tube; consequently, a variation in tube voltage changes the radiation dose and image quality. Reduction in tube voltage results in reduced radiation dose when all other parameters are held constant. This will increase the image noise and cause contrast changes. Several studies have demonstrated an ability to affect radiation dose and image quality by using a lower

tube voltage (Funama et al., 2005; Kalender et al., 2009). A current commercial technology called CARE kV from Siemens Medical Solutions (Forchheim, Germany) automatically adapts the tube voltage and tube current for each patient and clinical indication. The aim is to optimize the contrast-to-noise ratio (CNR) and minimize the absorbed dose.

2.2.3 Protection Of Radiosensitive Organs: ,

Such as the breast, eye lens, and gonads, is especially important in pediatric patients, adolescents, and young adults. However, the use of protective shields made of bismuth or other materials with a high atomic number over sensitive organs during CT investigations is controversial. The shields may cause streak and beam hardening artifacts, increase noise, and result in inaccurate CT numbers (Vollmar and Kalender, 2008). Organ-based TCM in which the tube current is reduced for a certain range of rotation was developed recently to protect radiosensitive organs from direct exposure. Wang et al. (2011a) concluded that the use

of organ-based TCM resulted in a similar reduction in the dose to the breast as achieved with bismuth shielding without affecting image noise or CT number accuracy.

The phenomenon of *over-scanning* is the exposure of tissue that is not reconstructed in topographic images (Hsieh, 2009). Due to reconstruction requirements, helical CT scans start and end beyond the region of reconstruction. As the X-ray beams in modern CT scanners become broader, more and more wasted radiation is delivered to the patient by over-scanning (Tzedakis et al., 2005). One solution to this issue is the use of dynamic collimator technique. Christner et al. (2010) showed considerable

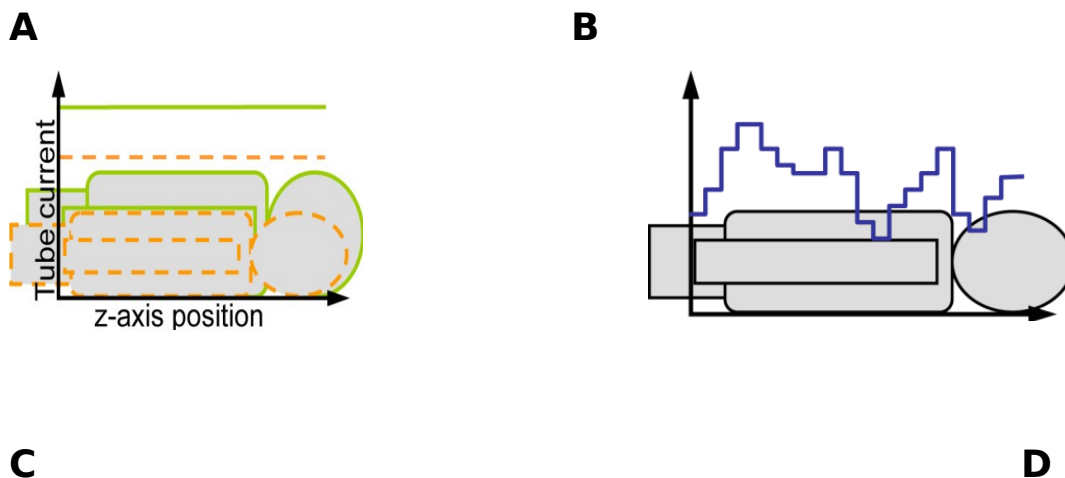
Dose reductions dependent on scanning length and examination using dynamically adjustable z-axis collimation. In conjunction with the use of lower doses, several attempts have been made to *filter images* to achieve noise reduction. Leander et al. (2010) showed significant improvement when using adaptive non-linear post-processing in adult abdominal CT. Similarly, Ledenius et al. (2010) found dose reductions of more than 10% for CT brain examinations in patients aged 6-10 years using an image-enhancing filter. However, post-processing image filtration has had reduced importance since the introduction of iterative reconstruction.

2.2.4 Automatic Exposure Control in CT:

In CT, automatic exposure control (AEC) automatically modulates the tube current in the x-y plane (angular

modulation), along the scanning direction (z-axis; longitudinal modulation), or both (combined modulation) (Kalra et al., 2005) (Figure 2.2). The modulation is performed according to the patient's size, shape, and the attenuation of the body parts being scanned.

The adaptation of the tube current is based on attenuation data from the localization radiograph and attenuation profiles or feedback from online measurements. AEC systems have a number of benefits: better control of the absorbed dose to the patient, improved consistency of image quality among patients, reduction of certain image artifacts, and reduced load on the X-ray tube, which increases its lifetime (Keat, 2005).



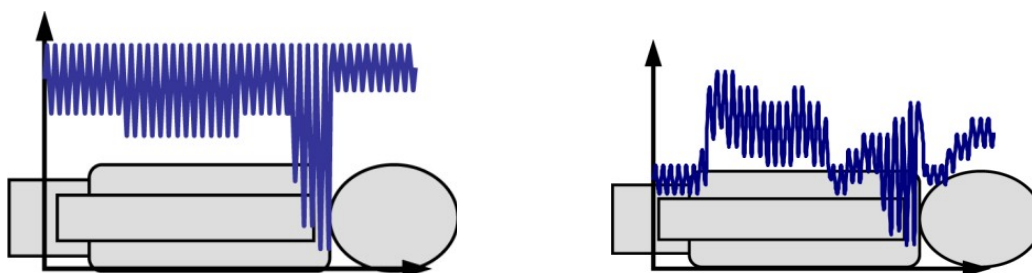


Figure 2.2 Illustrations of different AEC techniques in CT. Patient size AEC, lower tube current (mA) is used for a smaller patient (A), longitudinal AEC, lower mA is used for lower attenuating regions along the z-axis (B), angular AEC, based on asymmetry the mA is adjusted during the course of each rotation (C), combined AEC, a combination of the three techniques (D). Reproduced from Keat (2005) with permission from RSM Press.

All modern CT systems are delivered with AEC systems that modulate tube current in three dimensions. Each of these systems has different specifications and operates somewhat differently. However, the main principle is to manage the required image quality and radiation dose in a reproducible manner by adapting the tube current to the patient's size, shape, and attenuation.

2.3 Intravenous Contrast Medium:

Often, a need for contrast-enhanced CT examination exists in order to achieve a higher contrast between two different nearby structures. By using a contrast medium (CM), which

is commonly iodine based and administered intravenously or orally, the arteries, veins, tissues, and organs it courses through will be better visualized due to greater absorption and scattering of the X-rays. Several factors affect the contrast enhancement, which may be divided into three categories: patient (e.g., target organ, weight, and cardiac output), contrast medium (e.g., amount of CM, injection duration), and CT scanning (e.g., scan duration, scan delay, radiation) (Bae, 2010).

In addition, the radiation dose may be reduced if the CM achieves higher contrast between normal and diseased tissue (Watanbe et al., 2010).

2.4 Image Quality:

In a CT system, the linear attenuation coefficient (μ) is determined, which describes how the X-ray fluency rate is reduced by the object. The attenuation coefficient is presented as CT number relative to the attenuation of water. CT numbers (the signal) are given in Hounsfield units (HU) and, for an arbitrary tissue with attenuation coefficient μ_{tissue} , is defined as (Kalender, 2005):

$$\frac{\mu_{\text{tissue}} - \mu_{\text{water}}}{\mu_{\text{water}} - \mu_{\text{air}}} \times 100$$

water

$$CT \text{ number} = \frac{\mu_{\text{tissue}} - \mu_{\text{water}}}{\mu_{\text{water}} - \mu_{\text{air}}} \times 100 \text{ HU} \quad (2.6)$$

μ_{tissue}

Ideally, all pixel values would be zero when inserting a region of interest (ROI) in a homogenous water phantom. In reality, the values will be distributed around a mean value and the standard deviation (SD) is often used as a quantified measure of noise. Image noise has two main contributions in CT: quantum noise characterized by Poisson distribution and electronic noise that arises from the data acquisition system. However, SD is not a complete description of image noise as it provides no

information about the noise spatial characteristics, i.e. the noise can have different textures. A noise power spectrum reflects the degree of randomness at each spatial frequency and the shape reveals where the noise power is concentrated in frequency space (Dobbins, 2000).

The signal-to-noise ratio (SNR) is a description of the relationship between attenuation and image noise in a specified area. The difference in attenuation between adjacent structures, i.e. the contrast, has greater implications for the diagnostic use of images. The lower the contrast between two structures, the more their conspicuousness is reduced by the increased noise. This relationship is described by the contrast-to-noise ratio (CNR). In nuclear medicine the signal is characterized by the number of counts; fewer counts result in a higher noise level in the image. The CNR of lesion to background is essential for the detection of lesions in SPECT and PET images. Spatial

resolution refers to the ability of the system to depict variations in the distribution of radioactivity in the object. Because a limited spatial resolution volume is defined by the camera, collimator, radionuclide, acquisition protocol, and reconstruction method, the size of the lesion is also important. Below a certain volume, the reconstructed intensity tends to diffuse into neighboring voxels ‘spill-out’, resulting in low target-to-background ratios (underestimation of activity concentration in the target) (Hoffman et al., 1979). This effect is called the partial volume effect (PVE) or, as proposed by Skretting (2009), intensity diffusion (Figure 2.3). PVE may also involve another effect if the target is surrounded by background activity; the target signal will have a contamination component from the surrounding ‘spill-in’ (overestimation of the concentration of activity in the target) (Rousset and Zaidi, 2006).

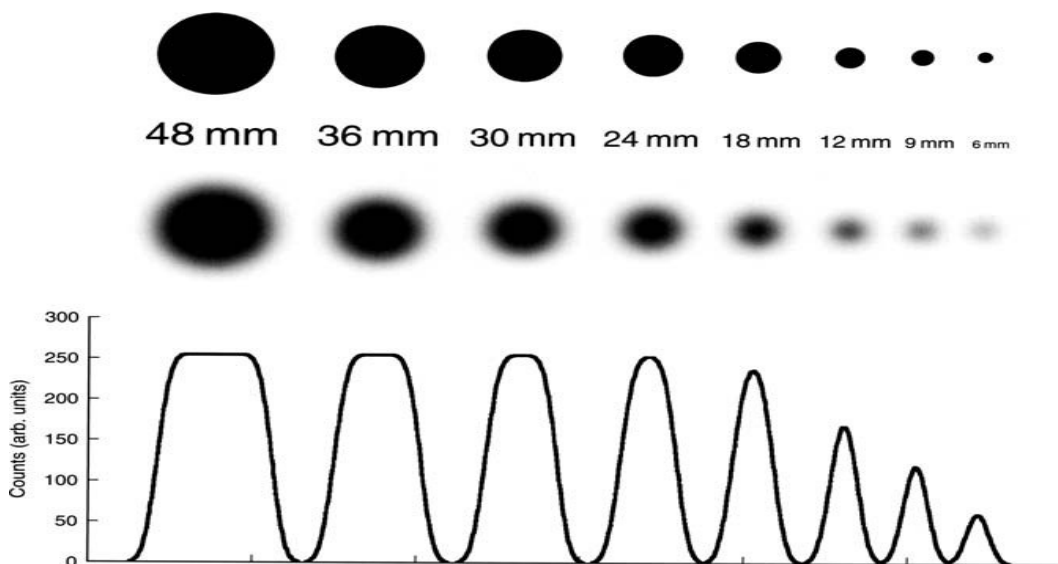


Figure 2.3 Illustration of PVE. The upper row shows cylinders of different diameters containing the same concentration of radionuclide. The middle row shows simulated SPECT images of the cylinders with an in-plane spatial resolution of 12 mm, full width at half maximum. One assumption is that the height of the cylinders is much greater than the axial resolution. The bottom row shows a profile through the centre of the images.

. Due to the PVE, the intensity decreases when the cylinder size approaches the resolution of the SPECT system. Reproduced from Cherry et al. (2003) with permission from Elsevier.

Several methods can be used to evaluate an imaging system (ICRU, 1996). Physical measures, such as detective quantum efficiency, take both the detector sensitivity and resolution properties into account and describe how an imaging system maintains the SNR (Bath, 2010). However, to predict the diagnostic potential and evaluate what sufficient image quality is for diagnosis in a specific examination, subjective evaluation is essential. Several types of observer performance studies have been used (Mansson, 2000). The choice of human observer study is dependent on the type of examination and conditions. Receiver operating characteristic (ROC) studies are appropriate when a specific

diagnostic task is investigated, i.e. when the task for the observer is to state pathological or normal findings (ICRU, 2008). Another approach is visual grading analysis (VGA), in which the reproduction or visibility of certain anatomical structures is assessed. VGA can be performed absolutely or relatively using one or several images as references. An underlying assumption is that pathological findings correlate with the reproduction of normal anatomical structures. An expanded VGA method was described by Bath and Mansson (2007) due to the often incorrect use of statistical methods when analyzing visual grading data. The method is termed visual grading characteristic (VGC) analysis and has been used in some studies, e.g., Carlander et al. (2008) and Leander et al. (2010). Another approach is visual grading regression, which is applicable when studying the effect from several factors (e.g. the kV and mAs settings) at once (Smedby and Fredrikson, 2010).