Radiation Safety and CT Dosimetry in PET/CT Imaging

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Contents

Abstract

Some of the particular radiation safety challenges associated with PET/CT are described for those more familiar with working in conventional nuclear medicine departments. Facility design and shielding calculations are described and examples given. Challenges in keeping doses to the staff and public down are described with some of the approaches to keeping doses as low as reasonably achievable. Computed tomography (CT) technology and dose metrics are described in some detail, and some guidance on quality control checks and patient dose in CT and PET/CT is given with references to sources of information on these topics in this rapidly changing field.

2.1 Introduction

Nuclear medicine departments have historically been designed and operated with consideration of the workflow through the department and local shielding of sources and radioactive material. Less consideration of any structural shielding within the facility has been made in the past.

For traditional nuclear medicine studies, radionuclides are generally administered in the

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department. Patients are then encouraged to leave the department during the uptake phase to both improve the uptake of the radiopharmaceutical and to minimise the radiation risk to the staff and the public from the gamma radiation emitted from the patient.

It was accepted practice for the staff to be in the same room as the patient during the scan on a gamma camera. The introduction of new technology in the form of SPECT/CT has caused a change to this approach, and it is now an accepted practice to have a separate shielded control room to protect the operator from the CT radiation scattered from the patient. It can also be seen from the whole-body monitoring data that this approach reduces staff exposure.

SPECT/CT has also required the nuclear medicine staff to learn new skills and acquire new knowledge around CT operation, dosimetry and quality control techniques. These will be covered later in this chapter.

PET/CT is different from conventional nuclear medicine. The energy of the gamma ray is higher, and patients spend longer in the department, thereby irradiating the staff to potentially significant levels.

The 511 keV gamma rays emitted as the electrons and positrons annihilate are more penetrating than the 140 keV gamma ray emitted by technetium-99 m (Tc-99m). The energy is the same for all radionuclides used, e.g. fluorine-18 (F-18) and carbon-11 (C-11). The half-lives of the different radionuclides can vary [1]. F-18 is most readily available and has a half-life of 110 min so that the radiopharmaceutical can be distributed some distance and travelling time from the point of manufacture. This means that PET/CT scanners can be installed in hospitals across a region with access to a single cyclotron.

The radionuclide is attached to a pharmaceutical as in conventional nuclear medicine. F-18 is commonly used as the radiolabel with FDG, although new agents are being introduced [2]. FDG is taken up in tissues in a similar way to glucose, and whilst it seemed a promising agent in neurological imaging, it is now widely used in tumour imaging for diagnosis, staging and monitoring of treatment.

As the uptake phase duration for this agent is typically 60 min and patients must rest in a quiet, warm environment to minimise uptake in muscles and then undergo a scan taking typically 20 min, the design of PET/CT facilities for F-18 FDG imaging has been a challenge to those used to conventional nuclear medicine.

This chapter will therefore consider some of the issues around this work that may be less familiar to those working in conventional nuclear medicine:

- Facility design
- Shielding calculations
- Staff exposure
- Optimisation of radiation protection in practice – fixtures fittings and accessories
- Optimisation of radiation protection in practice – staff dose
- CT technology
- Dosimetric quantities in CT
- Patient dose in CT
- CT quality control checks
- Patient dose in PET/CT

The patient flow through a PET/CT facility depends on the radiopharmaceutical, its form and means of administration (some are gaseous), the uptake phase, scan phase and discharge. The examples in this chapter will concentrate on F-18 FDG, but the principles can be extrapolated to other agents.

2.2 Facility Design

Standard building materials and hospital construction techniques may not provide sufficient protection for PET/CT facilities. The attenuation required may be greater than that afforded by brick or block construction, and scatter from the CT element of the test may require some shielding up to the underside of the soffit of the floor above; floors and ceilings may not afford

sufficient protection from either the 511 keV gamma rays or the CT radiation.

The design of any facility should involve many professionals who have expertise in their individual areas – architects, estate experts, structural engineers, mechanical and electrical engineers, patient representatives and project managers. Input from clinical staff similarly needs to be multidisciplinary so that the patients can be at the centre of the design but that workflow and radiation safety can also be optimised at the design stage.

Input from the equipment suppliers can also help and inform the process. A number of factors are required to be assessed to enable a design to be developed:

1. Workload – patient numbers and examination types. This will dictate the number of uptake bays.
2. Location – space is usually allocated or set by the boundaries of a building. This may limit the size of the facility. Space constraints are a common issue.
3. Potential future developments – consideration of these may be needed to enable new techniques to be adopted. For some facilities this may be less important, and an efficient throughput may be the overriding design consideration.
4. Equipment specification – this will inform the choice of scanner room size, floor loadings, electrical provision, etc. Quality control phantoms and sources should also be considered within this space.
5. Preparation room – a good ergonomic design will improve workflow and help with staff morale. This room may or may not include an area/uptake for waste and a spill kit.
6. Administration areas – privacy and dignity should be considered alongside the workflow and safety issues.
7. Consultation rooms might be considered. The patient consent process can be lengthy and might result in lower staff dose if this area is not close to the uptake areas.
8. Discharge – consideration might be required for patient refreshment after the scan or for those who need to wait for transport away from the facility. The patient will still have some residual activity and be a radiation hazard.
9. WC – a hot toilet will be required. Staff and/or visitor facilities should also be considered.
10. Other facilities – office space, staff rest areas and changing facilities, storage, etc. These are sometimes overlooked at the design stage and will be difficult to incorporate later.
11. Injectors – if injectors [3, 4] and/or automatic dispensers are to be or may be used, then consideration at the design stage is recommended.
12. Patient flow (see Fig. 2.1) – the processes that are undertaken as the patient moves through the facility, the situations when shielding or other operational procedures are needed to minimise dose can help with the design. Flow of radioactive material can be considered in a similar way.

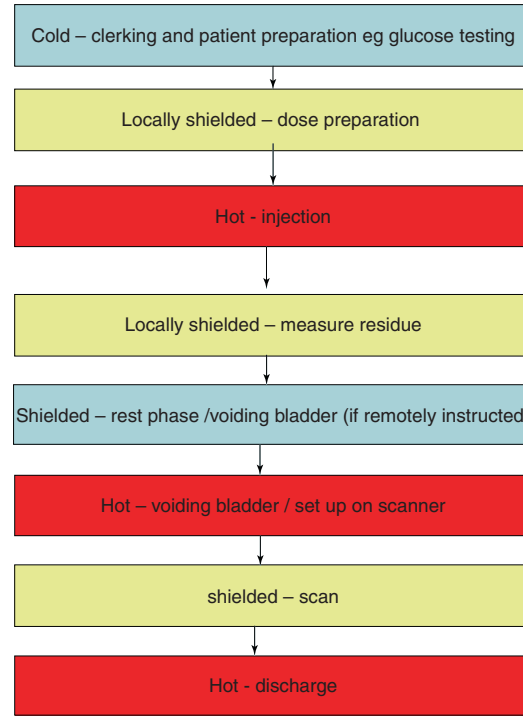


Fig. 2.1 Patient flow through the facility, highlighting hot, cold and shielded stages, and colour coded in proportion to the hazard

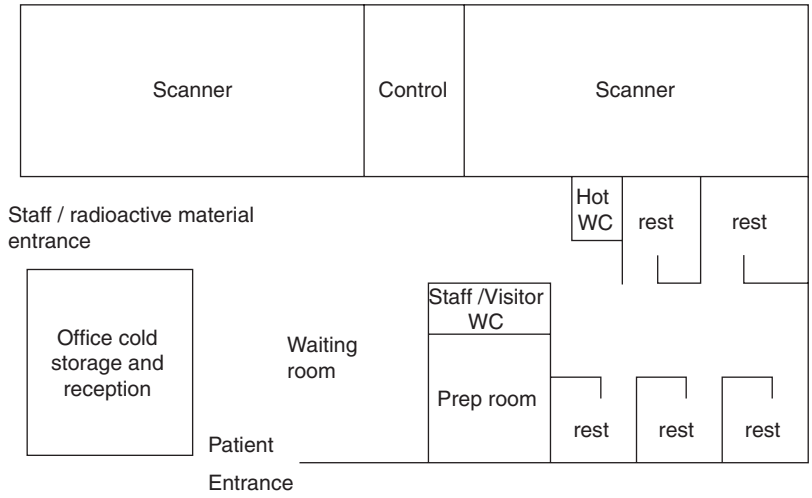


Fig. 2.2 An example of a well-designed facility

Consideration of the following points is recommended to minimise staff exposure and the cost of shielding materials:

1. Maximise the distance between hot patients and staff/members of the public.
2. Eliminate lines of sight between uptake bays and the operator console of the scanner (CCTV and intercoms can be built in cost-effectively at the design stage).
3. Use suitable local shielding for stock vials, syringes, waste, etc.
4. Consider the use of remote injectors and/or dispensers.
5. Consider the handling of QC sources.

The basic hazards to consider are:

- External dose rate hazard
- Contamination
- Emergency situations, e.g. a dropped vial

Standard radiation protection principles should be applied, i.e.:

- Distance
- Shielding
- Time

The inverse square law is very powerful and can dramatically reduce the level of shielding required.

Figure 2.2 shows a facility where there are no direct lines of sight between the rest bays and the control room where staff will spend most of their working time, i.e. there is always a barrier (which may be a wall) between the patient and the areas where staff spend significant periods of time. The preparation room is close to the rest bays and under supervision by staff in the control room and/or the office. The distance from the rest bays to high occupancy areas is maximised which will reduce the required shielding and minimise building costs. The hot WC(s) is/are close to the rest bays.

For maximum throughput with a single scanner, a minimum of three uptake bays are required. New scanners could increase throughput (or use reduced administered activity) which could impact on this choice.

Figure 2.3 shows a more challenging layout adopted on a number of mobile scanners, where staff in the control room are irradiated by patients in the uptake room and the scanner. The shielding required in this situation is much higher than for the facility in Fig. 2.2.

Further examples of design and layout are given in a number of publications [3, 5–8].

2.2.1 Shielding Calculations

Once the layout is established, shielding calculations are required to specify the thickness of

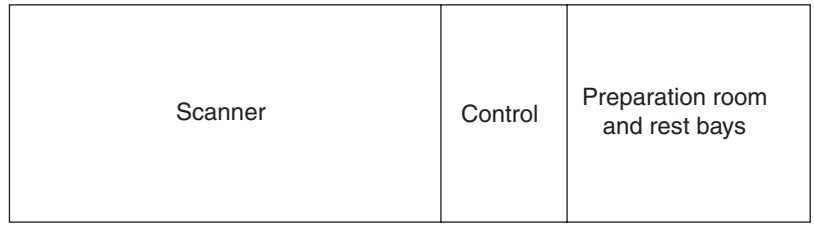


Fig. 2.3 Common layout for a mobile PET/CT scanner

Table 2.1 Limiting TVLs for lead and concrete

Material	Limiting TVL (mm)
Lead	15
Concrete (density 2350 kg/m^3)	150

Table 2.2 Typical transmission factors for building materials

Thickness and material	Transmission F-18
2.24 mm Pb	0.738
4½″ brick	0.03
9″ brick	0.09
100 mm breeze block	0.66
100 mm solid concrete block	0.27
200 mm poured concrete	0.046

barriers around the facility. The choice of building/shielding material may be clear – but should be established.

Common bricks and blocks may be the cheapest materials to use, although some contractors prefer poured concrete. Lead may be the only solution particularly for mobile vans. As some barriers may have a considerable thickness (up to 300 mm poured concrete), this may impact on the design/space requirements. The attenuation properties of common shielding materials can be found in AAPM Report 108 and the BIR report on shielding in diagnostic x-ray [5, 6]. A simple methodology using limiting tenth value layers (TVLs) is used below. The values for lead and concrete are shown in Table 2.1.

The TVL for other materials such as concrete blocks can be extrapolated using the ratio of physical density, e.g. 176 mm for concrete blocks with density 2000 kg/m^3.

Typical transmission factors for common materials are shown in Table 2.2.

The air kerma rate immediately postinjection of 370 MBq of F-18 FDG has been reported to be 45 μGyh^{-1} at 1 m [9]. There is some self-absorption within the patient of over 30 % from the air kerma rate quoted [1]. Table 2.3 shows the instantaneous dose rates through typical barriers and shielding materials in diagnostic facilities.

It should be noted that for planning purposes, a lower value is used to include a correction for radioactive decay and for excretion during the procedure.

The following examples of layouts and shielding calculations will concentrate on F-18-FDG scanning assuming an administration of 370 MBq, a rest phase of 1 h and a scan time of 20 min.

The following should be established to enable the calculations to be performed:

1. Workload – this will dictate the time a hazard is present in a particular area, e.g. uptake bays, scanner and discharge area.

   At worst all uptake bays and the scanner might be occupied 100 % of the time.
2. Dose constraint – this is typically 0.3 mSv per annum to members of the public and 1 mSv to staff. Note it may be difficult to achieve 1 mSv to the staff in a mobile design [10]. It should be noted that the staff will also receive dose from their contact with patients during injection, set-up and discharge. Total dose per annum for a high throughput of patients can approach 6 mSv, so it is important to minimise the dose that the staff receive through the facility design.
3. Calculation points – areas where staff or members of the public may spend time or become close to patients:
   - Control room
   - Offices
   - Rest rooms
   - Areas above or below the facility
   - Corridors
   - Clinic rooms

Table 2.3 Instantaneous dose rate at 1 m from an injected patient through typical barriers

Thickness and material	Instantaneous dose rate postinjection (μGy/h)
2.24 mm Pb	33.2
4½″ brick	13.7
9″ brick	4.1
100 mm breeze block	29.8
100 mm solid concrete block	12.2
200 mm poured concrete	2.1

Occupancy in these areas will need to be estimated. Some common occupancy factors are described in the BIR publication [5] with a range quoted to enable local knowledge of the use of the area to be applied. The minimum occupancy that is recommended to be applied is 5 % for car parks and other transiently occupied areas. Corridors may be between 10 and 20 %, but if it is known that a corridor is very rarely used, a factor of 5 % could be applied. Potential changes in the use of such areas in the future needs to be considered particularly if a low occupancy factor is applied. The distance between the source and the calculation points should be assessed.

The following dose rates can be used for planning purposes. They include integrating the initial instantaneous dose rate over the 1 h uptake phase and include a factor for evacuation of the bladder and decay over a 20 min scan [5]:

- 37 μGyh^{-1} at 1 m for the uptake phase
- 24 μGyh^{-1} at 1 m for the scan phase

The values are slightly more conservative than those quoted by the AAPM [6].

The barrier thickness is calculated by considering the critical points around a room such as the uptake bay. If the adjacent room is an office which is occupied 100 % of the time, and the critical point, e.g. the office chair, is 2 m from the centre of the rest bed:

- Using this distance from the patient – 2 m.
- Air kerma rate at 1 m from the patient is 37 μGy/h.
- The kerma rate corrected for distance is 9.25 μGy/h.
- The dose per annum (over 2000 h) is 18.5 mSv.
- The chosen constraint is 0.3 mSv.
- The attenuation factor (AF) or transmission required is no more than 0.3/18.5 or 0.02.
- The number of TVLs to attenuate to this factor is $\log_{10}$ (1/AF).
- Therefore, 1.79 TVLs are required, i.e. 270 mm concrete or 30 mm lead.

This can be repeated many times over to look at all critical points around the facility. Some centres choose to look at the worst-case point and apply that barrier thickness to the whole room to minimise the chance of mistakes during construction.

However, individuals are likely to be irradiated from more than one source, and the doses from all should be considered and summed [5]. If the approach above is used, the dose constraint could be divided between all sources, or an iterative approach using different wall thicknesses to share the dose burden more equally can be used. A number of authors have described different approaches [11–13].

The resultant required wall thickness can vary, but 300 mm concrete will almost always be adequate. This is to protect persons around the facility and will normally only be required to be a height of 2200 mm above finished floor level. Areas above and below may need some consideration.

CT scatter requires shielding to the underside of the soffit [5]. This might be most cost-effectively achieved by fitting a minimum of 1.3 mm lead above the walls to the soffit. The walls below this height will normally be thicker than this to protect from the gamma rays. Consideration needs to be given to air conditioning and other service access into the room. The penetrations can be large and often in the worst place from a radiation protection perspective, e.g. over the doors, which is often the place where scatter is highest.

2.2.2 Staff Exposure

Staff exposure is potentially high as can be seen by the shielding calculations above. The member of

the public dose limit is exceeded with 12 h at a metre from a patient during the uptake phase. Whole-body personal monitoring is recommended.

Finger dose can be exceptionally high with 500 mSv being reached within 3 h at a distance of 100 mm from a stock vial containing 10 GBq of activity. The inverse square law breaks down at closer distances, but it is clear that shielding and good technique are required. Finger dose monitoring is recommended for all staff manipulating radioactive sources in PET.

Eye doses may need to be considered in the light of the new dose limits [14] if whole-body doses are very high, but recent publications suggest that with adequate controls, dose limits should not be approached. However, radiation dose management in all hybrid imaging especially PET/CT can be a concern [15]. Direct reading dosemeters can be very useful devices to look at daily dose, to set alarms for staff who stand close to patients and can be used as an aid to audit.

2.2.3 Optimisation: Fixtures, Fittings and Accessories

Optimisation in terms of radiation protection is about keeping doses as low as reasonably achievable. The external dose rate hazard and the contamination risk both need to be considered. For the external dose rate hazard, the basic principles of operation in radiation protection apply:

- Distance
- Shielding
- Time

Distance is a powerful measure as the inverse square law comes into play. The basic design and layout can be used to maximise the distance between patients and staff, but there are times when staff need to be close to sources and the patient. The greatest hazard is from the stock vial which will need to be unpacked and measured. This could typically hold 10–80 GBq of activity. Simple handling tools can be used to maximise the distance between finger tips and the source

Fig. 2.4 Long-handled tool for handling stock vial

(Fig. 2.4). Long-handled forceps and other devices commonly used in nuclear medicine or radiotherapy can also be used.

The vial is generally placed into a dispensing unit which incorporates thick shielding (Fig. 2.5) and is designed to minimise exposure.

Automatic dispensing units are also commercially available and might be worth considering in high-throughput units. It is common practice to sink the dose calibrator into the preparation bench and to have a route to drop waste into a shielded waste bin. Bench top shields are recommended to protect the body and the eyes (Fig. 2.6).

All this local shielding is thicker than conventionally used in nuclear medicine. The weight may also need to be considered. Entry and exit portal monitors need careful consideration to eliminate background and to ensure contamination is not spread.

Plastic scintillators are used to detect beta emissions from the radionuclide although the gamma emissions are sufficient that some conventional sensitive scintillators may be adequate for contamination monitoring [16]. Storage and handling of sources used for quality control need

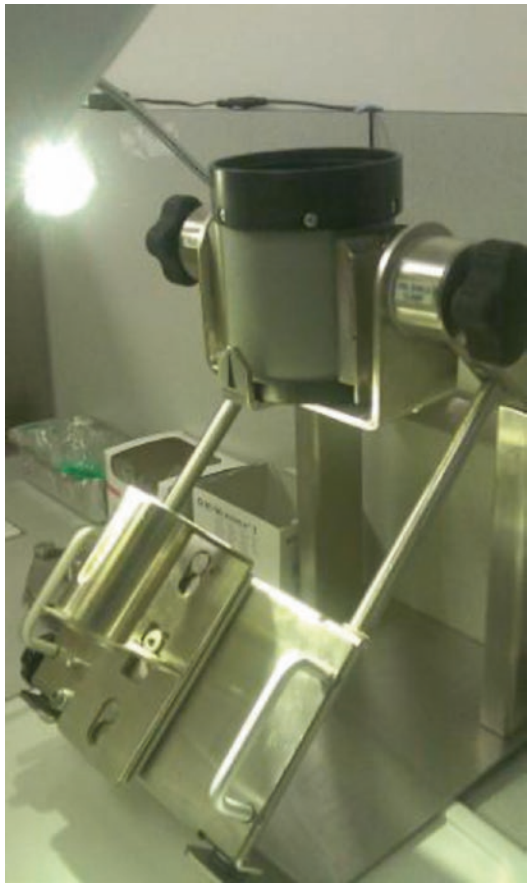


Fig. 2.5 Heavily shielded manual dispensing unit to hold the stock vial

to be considered. Sources should always be handled at the opposite end from where the activity is sited. Once doses have been dispensed, a syringe shield is required and the injection should be carried in a shielded box (Fig. 2.7).

There has been some debate as to whether Perspex or tungsten syringe shields reduce finger dose more effectively, but it is generally accepted that tungsten is more effective [17]. All syringe shields are bigger and more bulky than conventional shields, and staff should be initially trained to manipulate the radionuclide with inactive solutions. All surfaces must be easy to clean and decontaminate, flooring should be smooth without joints and cover up the wall and there needs to be space for decontamination kits and materials.

Risk assessments are essential to assess the level of protection required. High workload

Fig. 2.6 Dispensing unit with bench top shield, sunken calibrator and waste route into shielded cupboard

departments may need more automated accessories to keep doses low.

Contingency plans must be prepared for all obvious problems – fire, flood, theft and contamination. The action to be taken in the event of a stock vial being broken or a single injection being dropped should be known and understood by all staff involved. Contingency plans should be practised. All staff should be experienced in clearing spills, but for training purposes, inactive materials should be used.

2.3 Staff Dose and Optimisation

Once the service is operational, further optimisation of doses is likely to be possible as staff become familiar with processes and local practices.

If the total whole-body dose results for all staff are plotted over time against the number of patients through the system graphs, such in Fig. 2.8 might be observed. This shows that as the

Fig. 2.7 Syringe shields and carrying boxes for individual injections

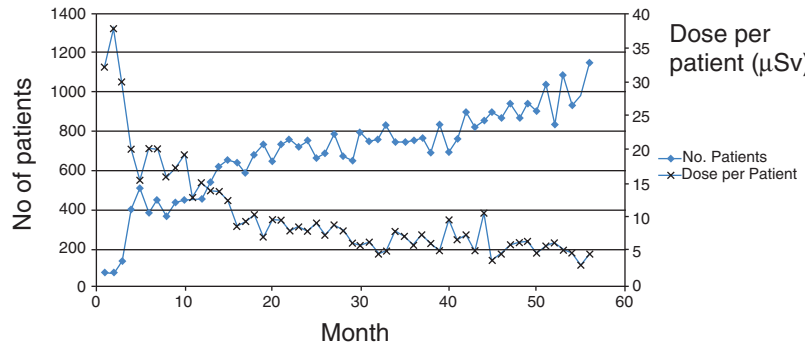


Fig. 2.8 Staff dose over time against increasing workload (Courtesy of InHealth Ltd)

workload increases, so does dose efficiency which plateaus out eventually.

The use of electronic personal dosemeters is advocated to monitor and help minimise dose for the whole procedure and individual parts of the procedure. A number of studies have been published showing the average dose per patient [18–22]. For a 370 MBq injection, the total dose per patient has been reported to be between 3.3 and 9.2 µSv.

Careful measurement of the doses recorded on electronic personal dosemeters can show the dose received during each stage of the process [22] as shown in Fig. 2.9.

Review of doses for individual members of staff can show that the doses vary by up to a factor of 5, giving further opportunity for optimisation and staff training. The opportunity to reduce these doses using automatic dispensers and injectors has been mooted [4]. There is some published data to suggest they can be successful, but it must be remembered they are in no way a substitute for excellent radiation protection in practice. Another tool advocated is to show areas where doses might be high within a facility – highlighting hot spots and the safest places to spend time as shown in Fig. 2.10 [22].

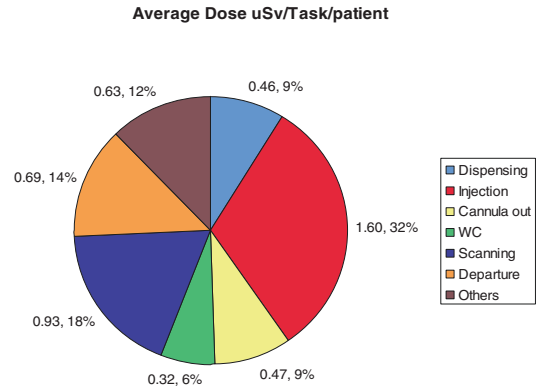


Fig. 2.9 Average dose for each part of the procedure (Republished with permission of British Institute of Radiology, from Peet et al. [22])

It should also be noted that optimisation is an ongoing process. Success in reducing doses and keeping them at that level requires constant vigilance. The modality whilst having exciting clinical potential needs particular care around staff and public safety – finger doses and whole-body doses can be high and could without care approach dose limits.

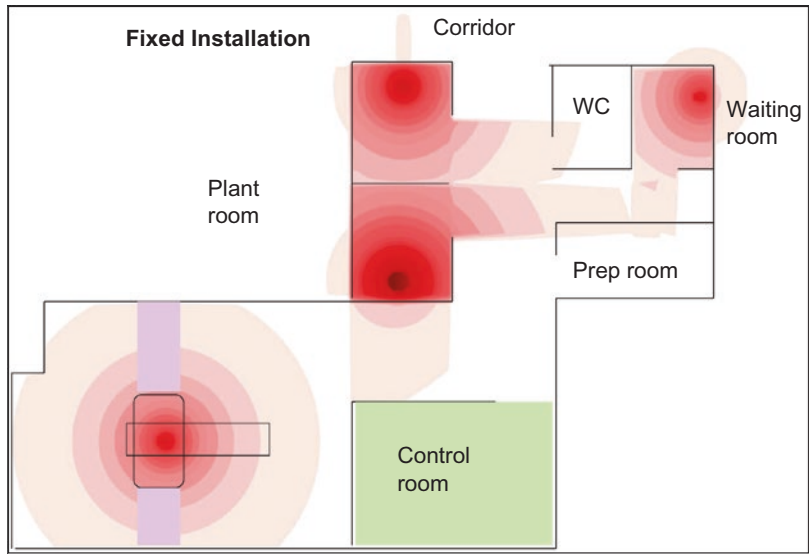


Fig. 2.10 Instantaneous dose rates plotted through the facility (Republished with permission of British Institute of Radiology, from Peet et al. [22])

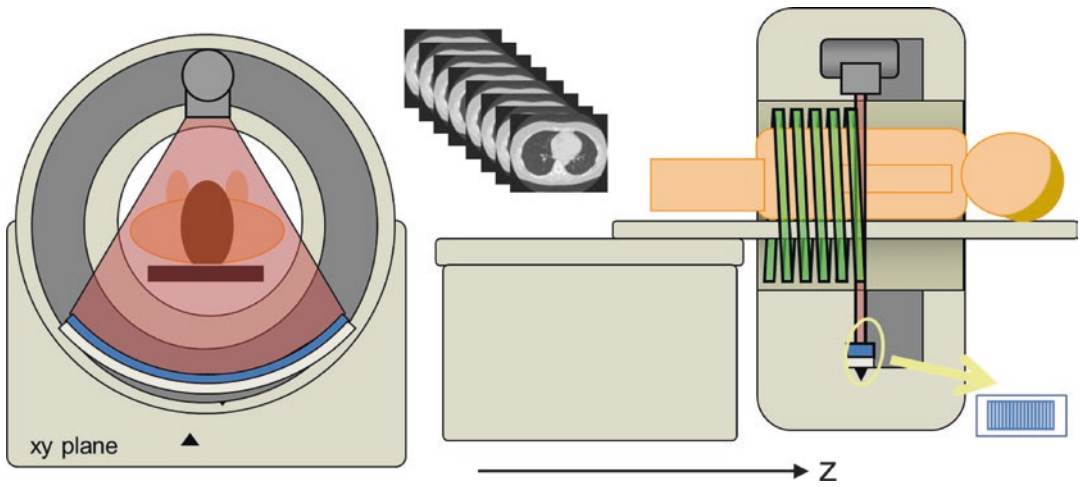


Fig. 2.11 Cross-sectional and lateral view of the CT scanner, with resultant images (Courtesy ImPACT)

2.4 CT Technology

2.4.1 Basic Technology

A CT scanner consists of a rotating gantry, a couch and an operator's console. The gantry, within an external housing, consists of an x-ray tube diametrically opposite to an arc of detectors, usually described as a detector row or rows. The whole construction rotates around a central point in space (the isocentre).

The patient lies on the couch whilst the x-ray tube and detectors are rotated around the patient, the x-ray source is activated for a scan and the detectors record the transmission of the x-ray beam through the patient. During this process, the couch can either be stationary, and the x-rays are switched off whilst the couch moves to the next position; this is called axial scanning. Alternatively the couch can be set to move continuously along the patient long axis, usually called the z-axis (Fig. 2.11), and this is called helical or spiral scanning.

The transmission data from the x-ray beam, attenuated by the patient, is collected by the detectors from many angles around the patient, and using either computerised 'back projection' or 'iterative' processes, a cross-sectional image of the patient is reconstructed.

The cross-sectional image presents maps of the x-ray attenuation properties of the tissues and organs. This can be used in PET/CT scanning to correct the PET image for the attenuation of the 511 keV gamma ray – known as attenuation scanning. This is in addition to the capability for anatomical localisation, as well as fully diagnostic anatomical information which can be acquired with or without contrast injection.

The exposure factors required for an accurate correction of attenuation are generally lower than those required for full diagnostic scanning where a less noisy image might be required. The dose the patient receives is related to these exposure factors, and this section describes the dose indices in common usage in CT and their relationship to effective dose (E).

2.4.2 Technology Developments

CT scanning has undergone many design changes since its inception. The introduction in 1985 of slip rings to transfer power and data, subsequently enabling helical scanning, could be regarded as a significant milestone and perhaps signals the design of modern scanners. In addition, since then, there have been numerous significant technological advances, and some of these need to be outlined to gain an understanding of the current dose indices.

It is not commonly remembered that the very first CT scanners in 1972 were dual-slice scanners, but this approach was very quickly dropped with a change in gantry design, and the single-slice design continued until 1991. Since then, over a period of 15 years, rapid developments in scanner technology have resulted in the transition from dual slice through 8, 16 and 64 'slice' and, in recent years, 320 detector row scanners.

Scanners are usually discussed, and sold, with respect to the competitive number of 'slices' that can be imaged in a single rotation. However, this can be extremely misleading. Any particular scanner model may reconstruct fewer (on older scanners) or more (on newer scanners) image slices per rotation than the number of detector rows. For example, an old 16-slice scanner may have 24 detector rows – the limitation at the time of their introduction being in terms of processing, cost of construction of thin detector elements and reconstruction capacity. Conversely a newer '64-slice' scanner may have 32 detector rows, a '640 slice' may have 320 rows or a '512 slice' 256 rows. This greater slice capability at a finer spacing than the actual detector rows is due to modern techniques in reconstruction and can be regarded as overlapping image slices. In terms of speed of coverage, therefore it is far more meaningful to categorise a scanner by the number of detector rows and the length of coverage along the patient axis by the whole detector bank, with consideration being given to the number of image slices acquired simultaneously (Fig. 2.12).

The time taken to undertake any particular examination scan is now shorter due to faster tube rotation times, longer detector arrays and more detector rows along the z-axis. Tube rotation times per revolution have been reduced to less than 0.5 s (of the order of 0.3 s to enable cardiac imaging). This together with the greater number of detector rows, and helical scanning, removes the need for breath-hold imaging in many circumstances. A large volume of the patient can therefore be scanned in a relatively short period of time (e.g. 20 s for the chest abdomen and pelvis). Radiation exposure must become a consideration both in terms of equipment design and operation of the scanner.

Image reconstruction has progressed from analytical filtered back projection methods to statistical iterative reconstruction techniques. On most modern scanners, both approaches are available. By the efficient use of the x-ray attenuation information, iterative techniques reduce image noise. Whilst this gives improved image quality, it also alternatively presents the operator an opportunity to lower the tube current (thus reducing the radiation dose), in order to restore the noise to the original level which had

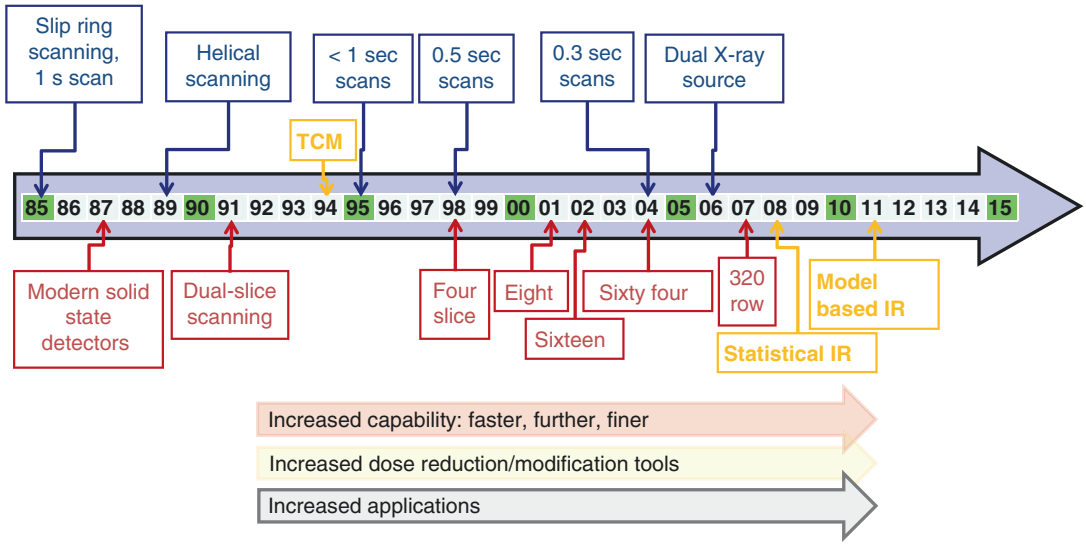


Fig. 2.12 CT technological advances 1985–2014

previously been accepted as suitable. A good description of CT technology and its evolution can be found in Kalendar's book [23].

2.4.3 Factors Affecting Dose in CT Scanning

The key parameters that affect patient (organ) dose are those that influence the photons delivered from the x-ray tube. Primarily therefore this is the tube voltage (kV), the tube current (mA) and the rotation time for one revolution(s).

There are also hardware features which are often built into the scan protocol, such as the x-ray beam-shaping filter which relates to patient body part (head or body) scanned (and sometimes patient size), and the x-ray tube focal spot which may be automatically adjusted according to tube current and imaged slice thickness.

In addition image reconstruction features may indirectly affect the dose. For example, the tube current may be set higher for a thinner reconstructed slice to allow enough photons for the required image quality. Reconstruction algorithms will affect the image noise, which also may require adjustment of the tube current.

The pitch in helical scanning may affect the dose, with a longer pitch reducing the average dose along the patient length. However, often the tube current is changed automatically to compensate.

The total dose imparted to the patient is governed by the site-specific (organ) dose, as well as by the scan length. A relevant aspect for consideration in helical scanning is that the total length of irradiation will be slightly longer than the resultant imaged volume, due to the interpolation of data in order to create planar images. This is of special relevance for shorter scan volumes, especially with wider beam widths, where the proportional increase in dose may be significant or where the end of the imaged volume is near organs of particular concern.

Most of the major manufacturers now have a 'dynamic collimation' that automatically, in real time, closes off the beam at the trailing and forward edges of the helical irradiation, i.e. at the beginning and end of the scan run, respectively. This ensures that unnecessary radiation is eliminated whilst keeping the appropriate transmission information in order to reconstruct the first and last images. Wider beam widths have the advantage of less penumbra proportionately and also faster scanner of a volume. Dynamic

collimators allow the use of wider beam widths without the penalty of extra irradiation at either end of the scan run.

Automatic exposure control (AEC) functions are available for all scanners. They operate by adjusting the tube current in order to attempt to match the tube x-ray output to the attenuation of the patient – adjusting according to the whole size, the relative rotational dimensions or along the patient long axis.

Each manufacturer initially developed one or more of these features, but now all have software packages to allow all three dimensions of automatic exposure control at once. These systems operate in different ways, either aiming to match the estimated patient image noise to a given noise value or adjusting the tube current compared to a reference patient size. To achieve this, either the attenuation data from one or two initial scan projection radiograph(s) is used or the attenuation information acquired by the detectors in the previous 180° rotation. Some allow the user to specify a maximum or minimum tube current in order to limit the extent of the modulation of the tube current.

The AEC features may increase or decrease the tube current depending on the set-up. On balance they are generally used to normalise the dose distribution to the patient, and also to reduce unnecessary dose, but they may not necessarily do this – depending on the specific way these features are used.

Tube voltage modulation is also available to adapt the tube voltage most suitably to the size of the patient, especially with iodine contrast agents where a lower tube voltage will produce a better contrast between tissue and the contrast agent. This feature as an automatic function (with selectable levels) is currently available from one manufacturer where the process can change both the tube voltage and the tube current. The dose will therefore be affected.

There are other features such as iterative reconstruction which do not affect the tube current and dose directly, but these may result in the use of a lower tube current to achieve the same image quality. Often iteratively reconstructed images can be produced at significantly reduced dose to achieve the same noise as filtered back projection images. Images constructed with iterative reconstruction may present a different texture to the noise than with filtered back projection and may also demonstrate different trends with changing tube current on the normal image quality performance parameters (noise and spatial resolution) than with filtered back projection.

The manufacturers have tended to develop two approaches to iterative reconstruction. There is usually a simple image-based iterative process which builds on the first filtered back projection image, and then more recently, others have been developed which operate on the raw data through modelling the focal spot and detector sizes, and these require more computing processes and can take longer.

All of these examples, of the effect of scan and reconstruction parameters on radiation dose, illustrate the need to record the exact scanner settings used in any given protocol when quoting dose values.

2.4.4 Dose Metrics in CT

A range of metrics are used in CT (as shown in Table 2.4) and require some understanding to enable manufacturer's information, and claims, to be understood and also to be able to make comments on their interpretation.

In CT a cross section of the whole body is irradiated in a single rotation, rather than at one angle and with an area limited by the collimator position as in projection radiography (Fig. 2.13).

Furthermore, the distribution of the dose is complicated by the rotation of the x-ray tube around, and movement of the table along, the *z*-axis, thereby irradiating a volume of tissue and organs within the patient.

Table 2.4 Standard generic dose metrics in CT

CT does parameters	
MSAD	Multiple scan average does
CTDI	Computed tomography dose index
DLP	Does length product
E	Effective dose
SSDE	Size-specific dose estimate

In the scan plane, for body scanning, the resultant dose at the centre of the irradiated volume may be half that at the periphery; however, the distribution tends to be more uniform in the head (Fig. 2.14).

Along the *z*-axis things are no less complicated. The dose profile for a single irradiation consists of the primary beam and scattered radiation beyond the nominal beam width (Fig. 2.15).

For an examination, whether with axial or helical scanning, these individual dose profiles combine to give a net irradiation profile along the *z*-axis. The extensive scattered tails of the single-slice dose profile contribute to give an average maximum value of the total volume dose profile which is usually greater than that of an individual slice, depending on the slice spacing. This is often called the multiple slice average dose (MSAD) (Fig. 2.16). Whilst this is not a commonly used term, it presents a very useful concept.

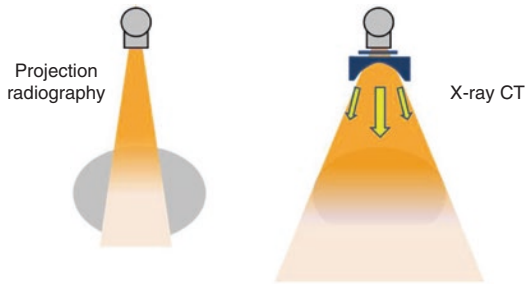


Fig. 2.13 Differences in dose distribution between conventional x-ray and CT – stationary beam (Courtesy ImPACT)

In helical scanning where the movement of the table, in one rotation of the tube, matches the beam width (pitch of 1), the dose pattern and transmission information are collected without any gaps along the *z*-axis. For faster relative table movement (a pitch greater than 1), or slower (a pitch less than 1), there will be subsequent gaps, or overlap, in the irradiation pattern along the *z*-axis. In all cases the data is interpolated to produce axial images.

The pitch also clearly has an impact on the average dose to the patient, with the average dose lower for higher pitch values, when other parameters such as the tube current remain the same. Many multi-slice scanners automatically adjust the tube current for changed pitch values, and in these situations, the dose remains constant.

The overall result of rotation and translation is a complex dose distribution as illustrated schematically in Fig. 2.17. The various dose indices have been developed to give simple approaches of measuring and expressing the average dose distribution.

2.4.4.1 Computed Tomography Dose Index (CTDI)

The CTDI is the standard dose index parameter used in CT. It is calculated from a single-slice measurement. It was originally designed to enable easy measurements to be made by a convenient method. Different forms of the CTDI have evolved with time and with scanner developments, and each depends on the specific use and application. A summary of the terminology for the current ones in use is given in Table 2.5.

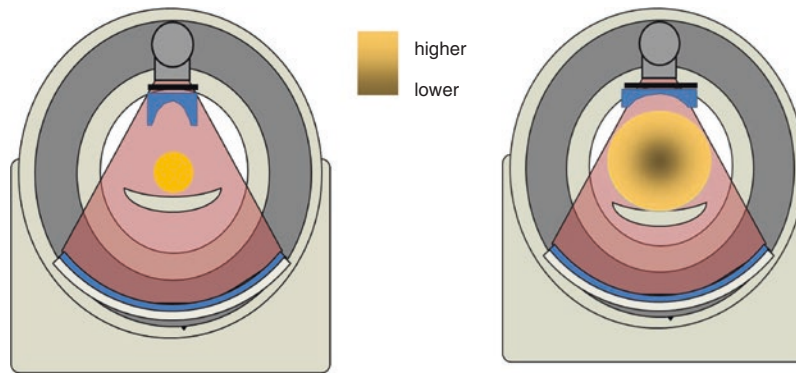


Fig. 2.14 Distribution of dose from a rotated beam – cross-sectional view. Periphery to centre ratio: head ~ 1:1, body ~ 2:1 (Courtesy ImPACT)

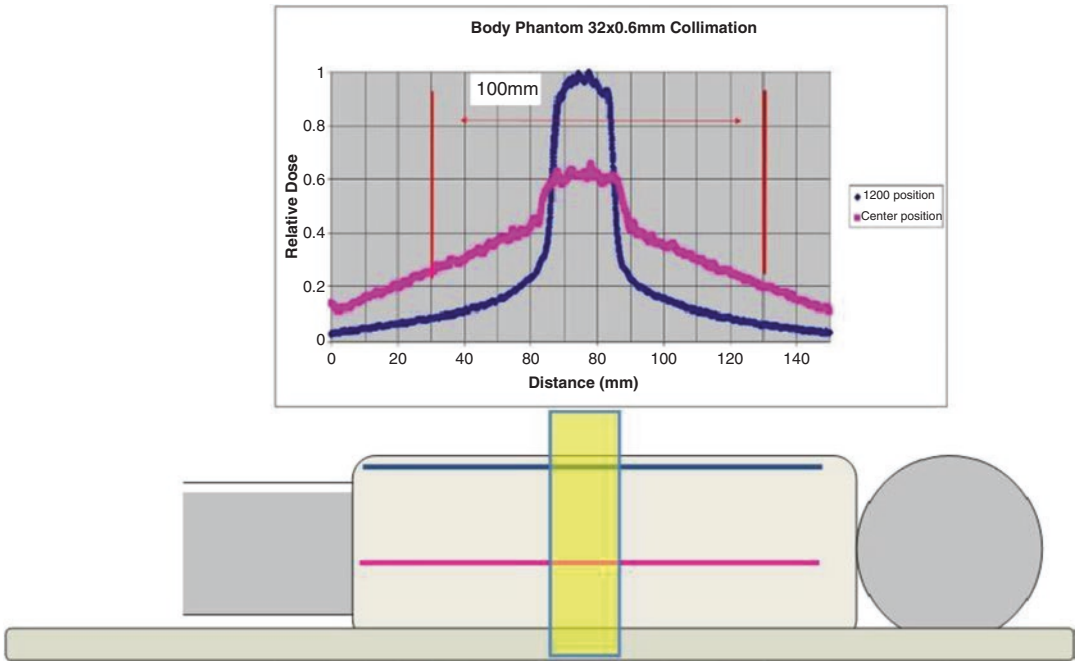


Fig. 2.15 Distribution of the single-slice dose profile along the *z*-axis, at the surface and centre of the patient or phantom (Courtesy ImPACT)

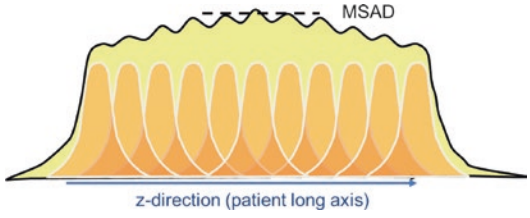


Fig. 2.16 A schematic view of an examination dose profile resulting from the contribution of scattered radiation from individual slices (Courtesy ImPACT). The MSAD (multiple scan average dose) is the value at the central region

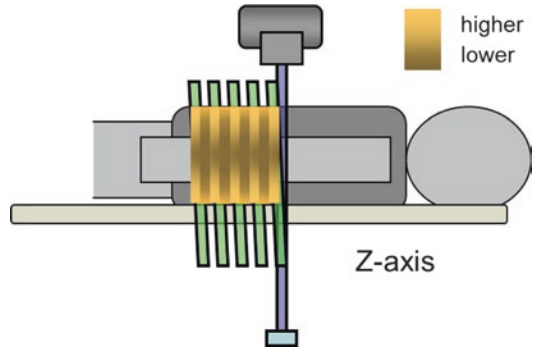


Fig. 2.17 Distribution of dose from a rotated beam along the *z*-axis: a complex radiation distribution (Courtesy ImPACT)

The general CTDI term consists of the integration of the single-slice profile, integrated over a given distance and essentially divided by the nominal beam width.

$CTDI_{100}$ is measured using a single rotation of the tube/detector, either in air or within standard polymethacrylate phantoms. The '100' in $CTDI_{100}$ refers to the length used from the single-slice dose profile. CT ionisation chambers for dose measurement are typically 100 mm, with a relatively small cross-sectional area (typically 1 cm^2).

The $CTDI_{100}$ is a calculation based on measuring the dose received to the chamber from the CT slice (including the primary and scattered beam) – essentially giving a value of dose that is averaged along the chamber length. By then dividing by the nominal beam width, it is essentially presenting an index whereby all the dose (primary and scattered) from the single slice is packaged within the nominal beam width (Fig. 2.18).

Though the CTDI is a calculation, based on the measurement from a single-slice irradiation, in concept it can be thought of as equivalent to the multiple slice average dose (MSAD), when measured in the same medium (air or phantom) and using the same length value. This is because both approaches take into account the scattered tails of the dose profile, the MSAD by measurement of the actual addition of scatter from neighbouring slices and the CTDI by the integration of dose profile, including scattered tails. Therefore, the $CTDI_{100}$ can also be interpreted as equivalent to the dose to the middle of a scanned length of 100 mm (Fig. 2.19).

Table 2.5 Standard generic dose metrics in CT

CTDI variations and terminology		
$CTDI_{100}$	CTDI one hundred	Calculated from a dose integral extending over 100 mm. Usually measured using a 100 mm pencil ionisation chamber
$CTDI_{air}$	CTDI in air	Measured in air
$CTDI_{w}$	Weighted CTDI	The weighted average of measurements at the periphery and centre of a standard phantom. Usually measured using a 100 mm pencil ionisation chamber
$CTDI_{vol}$	Volume-weighted $CTDI_{w}$	$CTDI_{w}$ adjusted for pitch in helical scanning
$CTDI_{\infty}$	CTDI infinity	Utilising the complete dose profile, including scattered radiation, integrated over 'infinite' distance

When the dose measurement is taken in air, the calculated $CTDI_{100}$ value is often referred to as $CTDI_{air}$, and this is very useful for quality control purposes. The weighted and volume CTDIs ($CTDI_{w}$ and $CTDI_{vol}$) result from measurements in standard-sized phantoms. They can also be used for quality control but are especially useful as indicators of relative patient dose when comparing protocols, scanners and values for similar-sized patients.

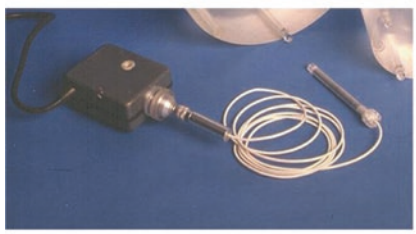
2.4.4.2 Weighted CTDI ($CTDI_{w}$)

$CTDI_{w}$ is calculated from $CTDI_{100}$ measurements undertaken in the standard 16 or 32 cm diameter polymethyl methacrylate (PMMA) phantom, which, being slightly longer than the standard CT ion chamber, is approximately 15 cm in length (Fig. 2.20).

The $CTDI_{100}$ is measured at the centre and periphery of the phantom to take into account the variation in dose across the phantom. This variation is more marked in the body as described earlier.

$$CTDI_{100} = \frac{1}{(N \times T)} \int_{-50}^{+50} D(z)\, dz$$

$$CTDI_{100} = \frac{\text{integral dose over 100 mm}}{\text{nominal beam width}}$$



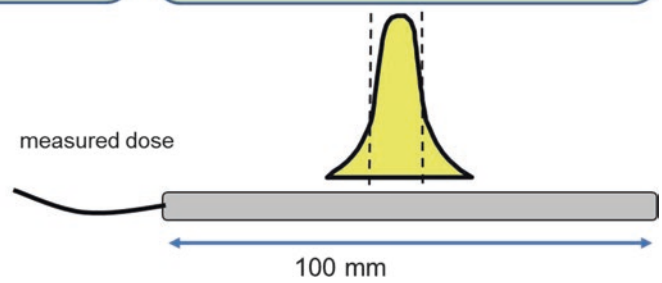


Fig. 2.18 The CTDI equation and presentation of an easy approach to remember it. D(*z*) is the single-slice dose profile, *N* is the number of image slices of thickness (*T*) that are acquired simultaneously and (*N*×*T*) is usually equivalent to the nominal beam width. Also shown are the CT ionisation chamber (*left*) and an illustration (*right*) of a dose profile from a single slice including the scattered tails and the chamber. The dotted lines represent the nominal beam width (Courtesy ImPACT)

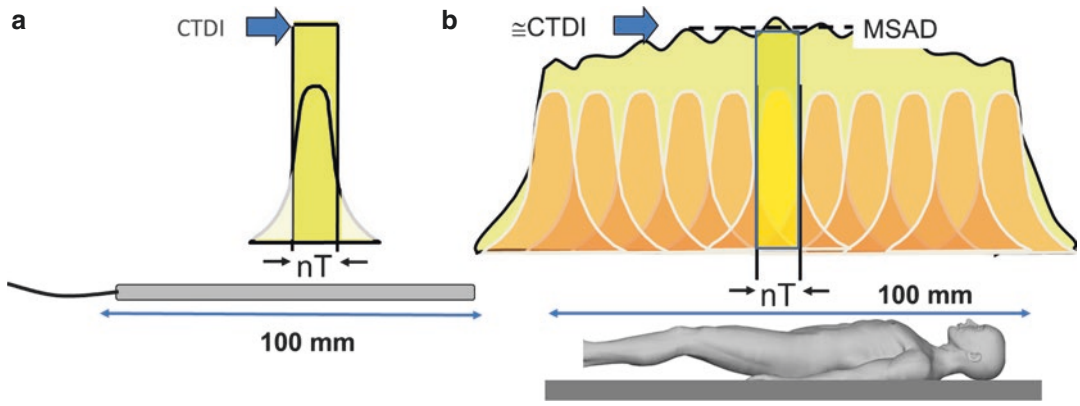


Fig. 2.19 Demonstrating the equivalence between the CTDI (**a**) and the MSAD (**b**) (Courtesy ImPACT)

Table 2.6 Conversion factors *K*, effective dose to DLP

Region of body	Conversion factor $K = E/DLP$ (mSv mGy^{-1} cm^{-1})
Head and neck	0.0031
Head	0.0021
Neck	0.0059
Chest[a]	0.014
Abdomen and pelvis	0.015
Trunk	0.015

Ref. 2004 CT Quality Criteria (MSCT 2004) [24], also found at AAPM Report 96 [25]

[a]The chest value is a factor that must especially used with caution as it is sometimes quoted as 0.017

The definition is given as

$$CTDI_w = 1/3 CTDI_{100,c} + 2/3 CTDI_{100,p} \quad (2.1)$$

This gives a weighted average for the cross section of the phantom.

2.4.4.3 Volume CTDI ($CTDI_{vol}$)

Since the CTDI is a calculated dose index from a single-slice measurement, a correction must be applied to make it relevant for helical scanning. For example, where the helix of the primary dose beam is spread out, the resultant net dose will be averaged to be lower (unless other factors such as the tube current are adjusted to compensate). This spreading out of the helix is called the pitch and is defined by the relative table movement of the couch per tube rotation with respect to the nominal beam width:

$$\text{Pitch} = \text{table travel per rotation} / \text{nominal beam width} \quad (2.2)$$

A correction for the pitch is applied to the $CTDI_w$ to give the volume CTDI ($CTDI_{vol}$):

$$CTDI_{vol} = CTDI_w / \text{pitch} \quad (2.3)$$

A protocol with a pitch of 1 and a given $CTDI_w$ and $CTDI_{vol}$ would result in a halved $CTDI_{vol}$ for a pitch of 2, providing no other scan parameters were changed (Fig. 2.21).

The $CTDI_{vol}$ (or the $CTDI_w$ with older scanners) is presented on the operator console of the CT scanner (Fig. 2.22) as described in IEC 60601-2-44 3rd edition. The values will appear both prior to the scan as well as afterwards to reflect the actual value. This is particularly relevant where automatic exposure control has been used. In these cases either the maximum or the average value will be shown for the scan.

The $CTDI_{vol}$ values will also be given in the dose report, which presented as an image with the scan, and be populated into DICOM header information that will go to the PAC system. For those scanners that also support the DICOM Radiation Dose Structured Report (RDSR), more associated details are available with the $CTDI_{vol}$ data.

$CTDI_{vol}$ values are used for establishing local and national diagnostic reference levels (DRLs) as described in International Commission on Radiological Protection (ICRP) 105 and the UK

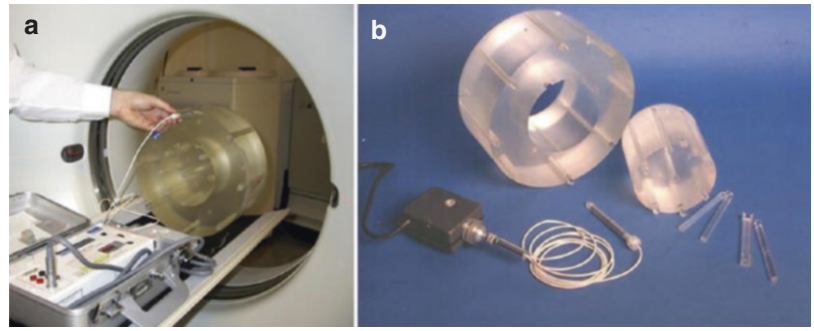


Fig. 2.20 Standard PMMA phantoms for the measurement of the CTDI. The head phantom fits inside the body annulus in this example of the phantoms. The standard CT ionisation chamber is also shown (Courtesy ImPACT)

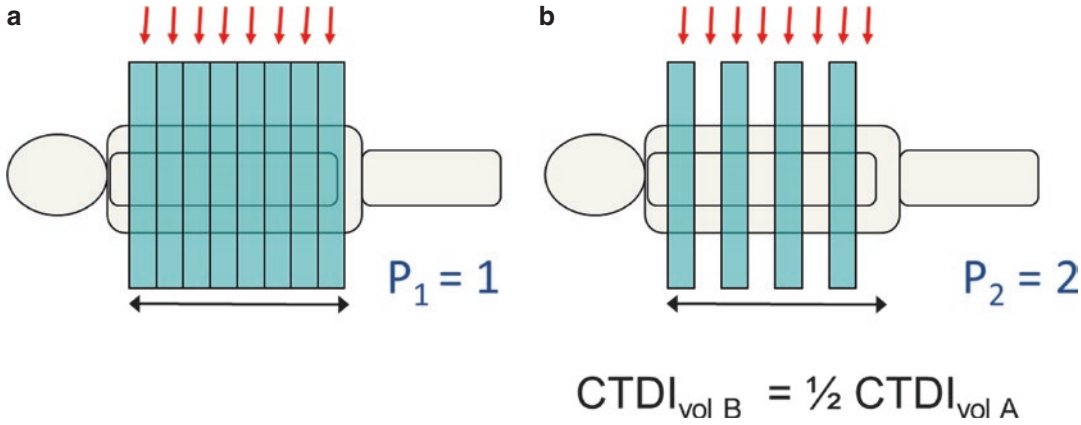


Fig. 2.21 Illustration of the effect of pitch on the calculation of $CTDI_{vol}$ (Courtesy ImPACT)

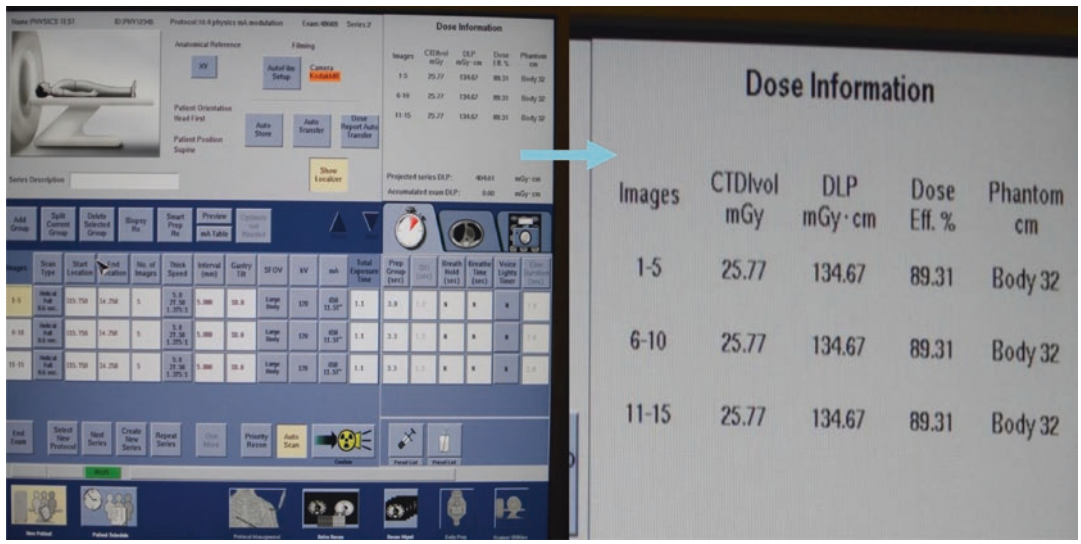


Fig. 2.22 Example of the presentation of the $CTDI_{vol}$ on the scanner console

IRMER regulations [27, 28]. DRLs will be discussed in more detail later in this chapter.

2.4.4.4 Dose Length Product (DLP)

Another metric can also be used to indicate dose taking account of the length of the scan – the dose length product. By utilising the $CTDI_{vol}$ as an indicator of 'dose per slice' and by taking into account the actual length scanned, a value can be calculated as an indicator to represent the total dose for the scan. This can then be considered to relate to radiation risk:

$$DLP = CTDI_{vol} \times L\,(\text{mGy.cm}) \qquad (2.4)$$

L is the scanned length in centimetres.

It can easily be seen that a scan that is twice the length of another results in a doubling of the DLP (Fig. 2.23), where all other parameters are the same.

DLP values are used, in addition to the $CTDI_{vol}$, for establishing local and national DRLs [27, 28]. They can also be used to convert to an approximate effective dose for a given body region as described later in this chapter.

2.4.4.5 The Measurement of $CTDI_{vol}$ for Wide-Beam Scanners

Wide-beam scanners, with beams greater than 80 mm along the patient length, i.e. the z-direction (Fig. 2.24), present a problem for the practical measurement of the $CTDI_{100}$ (and thence the calculation of $CTDI_{vol}$) as they have a beam whose primary beam is close to, or exceeds, the length of the CT ionisation chamber length of 100 cm.

An IEC standard (Edition 3, Amendment 1) [26] has addressed this in a pragmatic approach, by redefining $CTDI_{vol}$ for beams greater than 40 mm. This approach ensures that the calculated $CTDI_{vol}$ is kept constant with all beam widths,

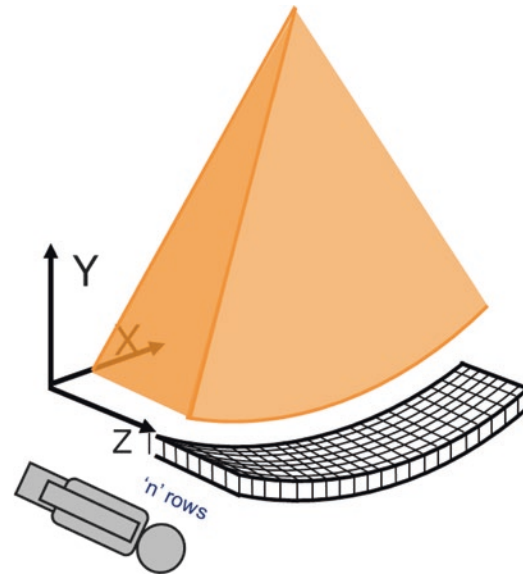


Fig. 2.24 Schematic view and photograph of the wide or cone beam and the axis orientation (Courtesy ImPACT)

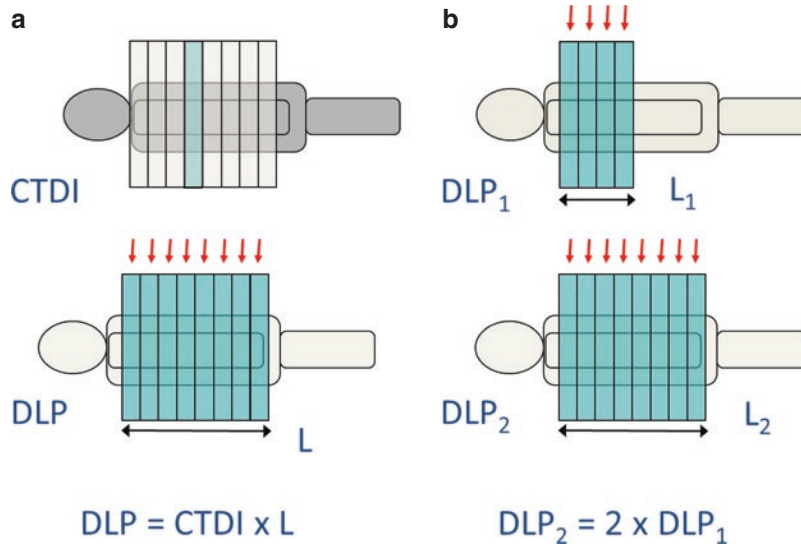


Fig. 2.23 Schematic view to demonstrate DLP and CTDI. In (**a**) the DLP is the length multiplied by the $CTDI_{vol}$. In (**b**) the DLP for length 2 (L_2) is twice that for scanned length 1 (L_1), whilst the CTDI is the same

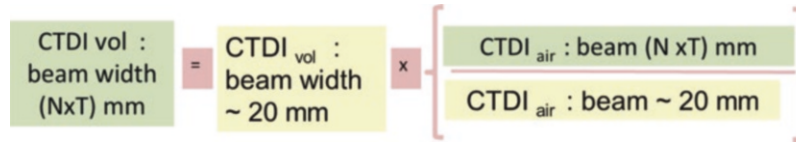


Fig. 2.25 IEC redefinition of $CTDI_{vol}$ for beams greater than 40 mm

despite the measurement tool (the CT ionisation chamber) being only 100 mm. This might seem an obvious approach for beams greater than 80 mm or indeed 100 mm; however, there are significant scatter considerations that also affect narrower beams; hence, the limit of 40 mm was chosen. The basic principle is to first undertake the measurements and calculations to obtain a $CTDI_{vol}$ for a suitable beam width of about 40 mm. A correction is then applied, based on the ratios of the $CTDI_{air}$ for the wide-beam width, relative to that at 40 mm (Fig. 2.25). $CTDI_{air}$ is easier to measure for wide beams due to the lack of scattered radiation, and this approach can be undertaken with existing equipment.

This pragmatic approach is clearly presented, with practical application in mind, in the IAEA report on wide-beam CT scanners [29], and further explored and presented in a paper in the British Journal of Radiology by Platten et al. [30].

These references also include some practical advice as to how to measure the integration of the dose profile (and thence the calculation of the $CTDI_{air}$).

2.4.4.6 Beyond $CTDI_{vol}$: Size-Specific Dose Estimate (SSDE)

The $CTDI_w$ and $CTDI_{vol}$ are dose indicators, and not patient dose values. They are highly valuable when used appropriately, such as comparing doses for different scan protocols and scanners as well as for establishing and comparing to National Diagnostic Reference Levels (NDRLs) which are based on standard patient sizes [28, 31].

However, these parameters cannot be used to compare actual patient scans unless the patients are of similar size. Patients come in many shapes and sizes, and the CTDI values only relate to a PMMA phantom of a fixed size.

Scanner settings generally should be adjusted to account for patient size, in order to give adequate dose to the detectors and thence a suitable image quality. This can happen either manually, through preset protocols, or on modern scanners with the use of automatic exposure control systems that adjust to meet a specified image quality.

In the scenario where the tube current has not been adjusted, the actual dose to a small patient, or child, would be greater than for a standard patient. However, the scanner would indicate the same phantom $CTDI_{vol}$ (Fig. 2.26). Similarly for a large patient, the actual dose would be less than for a standard-sized patient, but the scanner-indicated $CTDI_{vol}$ would be the same.

Conversely, by utilising tube current adjustment, it may mean that scans of two patients of different sizes, both resulting in similar image quality, will give an indication that one patient has a much higher $CTDI_{vol}$. The assumption might be that the larger patient received much higher dose; however, the actual dose delivered to the larger patient may be comparable to that of the smaller patient.

The AAPM have published two reports (AAPM 204 and 220 [32, 33]) giving correction factors that can be applied to the $CTDI_{vol}$ to give a dose indicator that takes into account the size of the scanned patient (Fig. 2.27). This revised dose metric is called the size-specific dose estimate (SSDE).

These correction factors were based on studies carried out by four different centres, using approaches based on measurements and Monte Carlo calculations on different sizes of 'CTDI'-type phantoms, elliptical anthropomorphic phantoms and voxel phantoms.

2.4.4.7 Beyond $CTDI_{vol}$: Scanned Length and Equilibrium Dose

$CTDI_{vol}$ is a dose index and as such is extremely valuable for quality control, scanner and protocol comparisons, as well as referencing to national

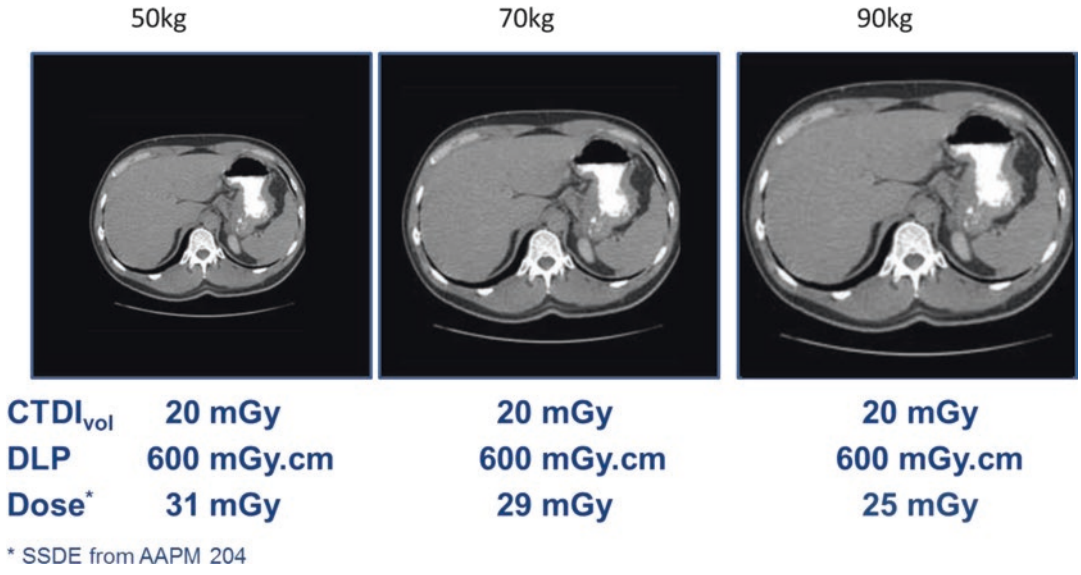


Fig. 2.26 Comparisons of $CTDI_{vol}$ (as measured for the 32 cm phantom) but presented for scans of different-sized patients and conversely an estimation of the actual relative dose received (using AAPM report 204)

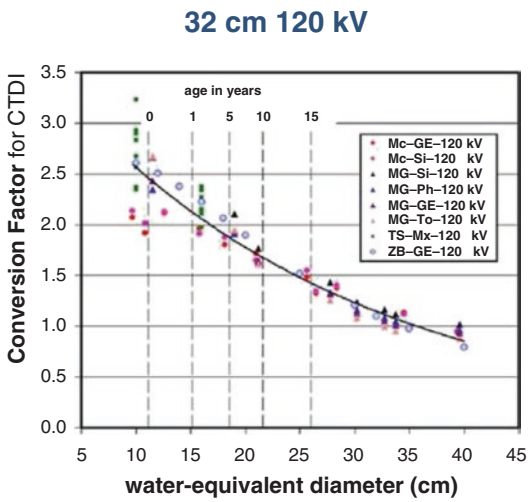


Fig. 2.27 Conversion factors based on the water equivalent diameter to apply to the $CTDI_{vol}$ to give an estimate of the SSDE (From AAPM report 204)

dose reference levels (National Diagnostic Reference Levels, NDRLS), as discussed later in this chapter.

However in a similar way to the endeavour to find a dose index that reflects patient size, there have been similar explorations to determine the relevance of the $CTDI_{vol}$ to the actual scanned length of an examination. The $CTDI_{vol}$ only utilises the scattered radiation over a total length of 100 mm, whereas the scattered tails of the single-slice dose profile have been shown to extend over a much greater distance. The term $CTDI_{infinity}$ is used for the limiting value of the generalised CTDI value where all the scattered radiation is considered. The issues are described in ICRU 87 [34] and AAPM 111 [35] and are likely to result in new phantoms and measurement methodologies in the future.

2.4.4.8 Effective Dose

Most people find the concept of effective dose the simplest as it enables a comparison of different scans and imaging modalities. However, it has to be used with caution and consideration given to derivation of values given.

Effective dose is based on utilising the radiation dose given to all organs in the body. Organ doses are calculated and weighted according to the sensitivity to radiation in terms of their statistical potential for the development of cancer (Fig. 2.28). The weighted doses are subsequently summed together. The resultant value gives an effective whole-body dose (effective dose, E). This value can be interpreted as the overall

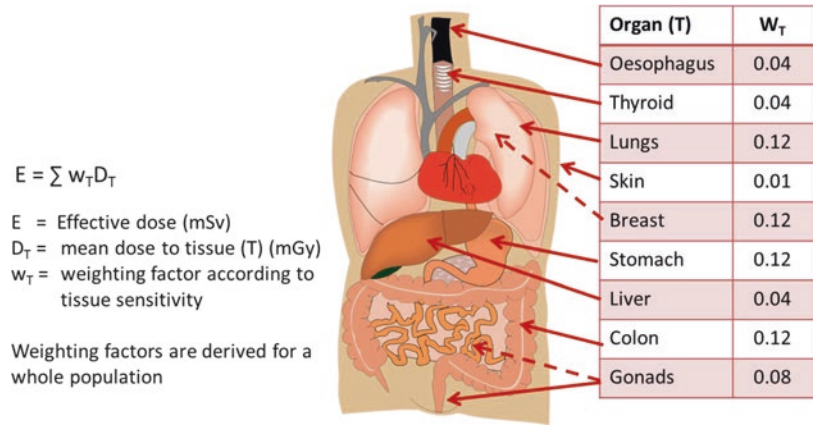


Fig. 2.28 Organ weighting factors for effective dose (Factors from ICRP 103)

potential risk of cancer development according to ICRP [31].

The organ weighting factors for the current ICRP report (2007) [31] have been updated with the major changes relating to the breast values (which have increased) and the gonad values (which have decreased).

Effective doses, and their associated risk estimates, are designed for a population and therefore do not take into account risk adjustments for age and gender. Effective dose should not be applied to individuals; if a situation requires some estimate of effective dose, then it must be used with caution, since it was not developed for this purpose.

When quoting effective dose (E), it is very important that the source of the weighting factors used is noted, whether the previous factors in ICRP 60 [30] or the current factors in ICRP 103 [31]. The phantom on which the calculation is made is also important. There are a number of phantoms that are widely used, and the ICRP has specified adult phantoms in their publication 110 [36]. It is not possible to make sensible comparisons of effective dose without this knowledge. If not noted, for example, this might be thought of as the equivalent of comparing two objects of the same length, one with a length value in centimetres and the other in inches. The numbers will be different, but they represent the same length of object.

The level of accuracy must also be borne in mind with these values, and therefore it may be possible to only consider broad estimates.

The impact of evolving International Commission on Radiological Protection (ICRP) recommendations concerning calculations of effective dose (E) are investigated by Shrimpton et al. [37], and their paper compares updated typical UK values for common CT examinations with previous data.

Organ doses, and thence effective dose, can be calculated using Monte Carlo techniques and with specific mathematical numerical or voxel (or hybrid) phantoms (Fig. 2.29).

Monte Carlo calculations are extensive calculations, where the x-ray source is mathematically described and used as the source for millions of mathematical generated photons. The path for each photon is followed through a statistical process with respect to its interactions with tissue. This process requires a number of elements, specialist knowledge of the x-ray tube and filtration, numerical or voxel versions of the phantoms, a Monte Carlo calculation package and often high computer processing power and/or time. There are many papers written in this field; a good overview paper is given in the references [24].

Effective dose can also be estimated from E/DLP factors (K-factors). These are specific factors which take the DLP value for specific regions of the body and where the effective dose has been calculated for the same scan parameters and body region.

There are a number of published sources. Sometimes the values are different depending on the examination region, the phantom and/or the ICRP report from which the underlying organ

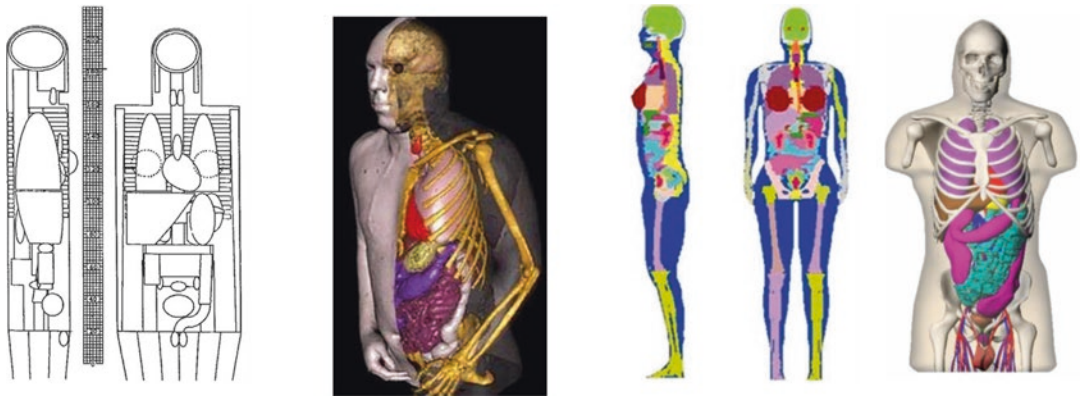


Fig. 2.29 A variety of phantoms used for Monte Carlo calculations

weighting factors are used. An example of published data is given in the 2004 CT Quality Criteria [25] (MSCT 2004) at http://www.msct.eu/CT_Quality_Criteria.htm# and the same values presented in AAPM 96 [38]. The chest value is a factor that must especially be used with caution as it is sometimes quoted as 0.017, compared to 0.014 in these references (Table 2.6).

Organ doses, DLP and effective dose, for specific examinations and scanner models, can also be calculated using commercially, or freely, available CT dosimetry software. One such is the ImPACT CT Calculator [39]. Others are also available [40, 41]. These utilise published organ dose data sets generated from Monte Carlo calculations. The examination and scanner can be selected easily, and conversions to adjust for newer scanner models can be made (Fig. 2.30).

2.4.5 Diagnostic Reference Levels (DRLs) and Optimisation in CT

DRLs are used as indicators of reasonable practice and are expressed in units of DLP or $CTDI_{vol}$. Much work has been done on assessing doses within CT. For example, the Public Health England Centre for Chemicals, Radiation and Environmental Hazards (CRCE), and its former affiliations (the Health Protection Agency, HPA and the National Radiation Protection Board, NRPB), has periodically published snapshots of CT scanner dose data from surveys of CT scanner practice in the UK. These surveys result in the quotation of typical $CTDI_{vol}$ and DLP for certain clinical examinations, as national reference doses, which are then formally adopted as the UK National Diagnostic Reference Levels (NDRLS).

The latest report, Doses from Computed Tomography (CT) Examinations in the UK (2011 review), was published in 2014 [42]. With repeat studies, these can show trends over time. Future national surveys should become easier with the introduction of automatic dose data collection systems; however, they can then present other difficulties due to the large volume of data.

The published data can help those working in PET/CT to see the range of doses for particular diagnostic examinations. Doses for attenuation scans are not as readily available from surveys such as these, but there is much information published from within individual centres particularly concentrating on optimising the scanning protocols used as described in the section below.

2.4.5.1 Optimisation in CT

Dose reduction has been discussed by Iball et al. [43] and can be used to optimise exposure factors in Table 2.7, to reduce doses without compromising the level of noise within an image. These concentrate on reducing the DLP and $CTDI_{vol}$ on individual scanners by understanding the functionality of the automatic exposure control (AEC) system in use on the CT scanner and by reviewing dose length product and image noise.

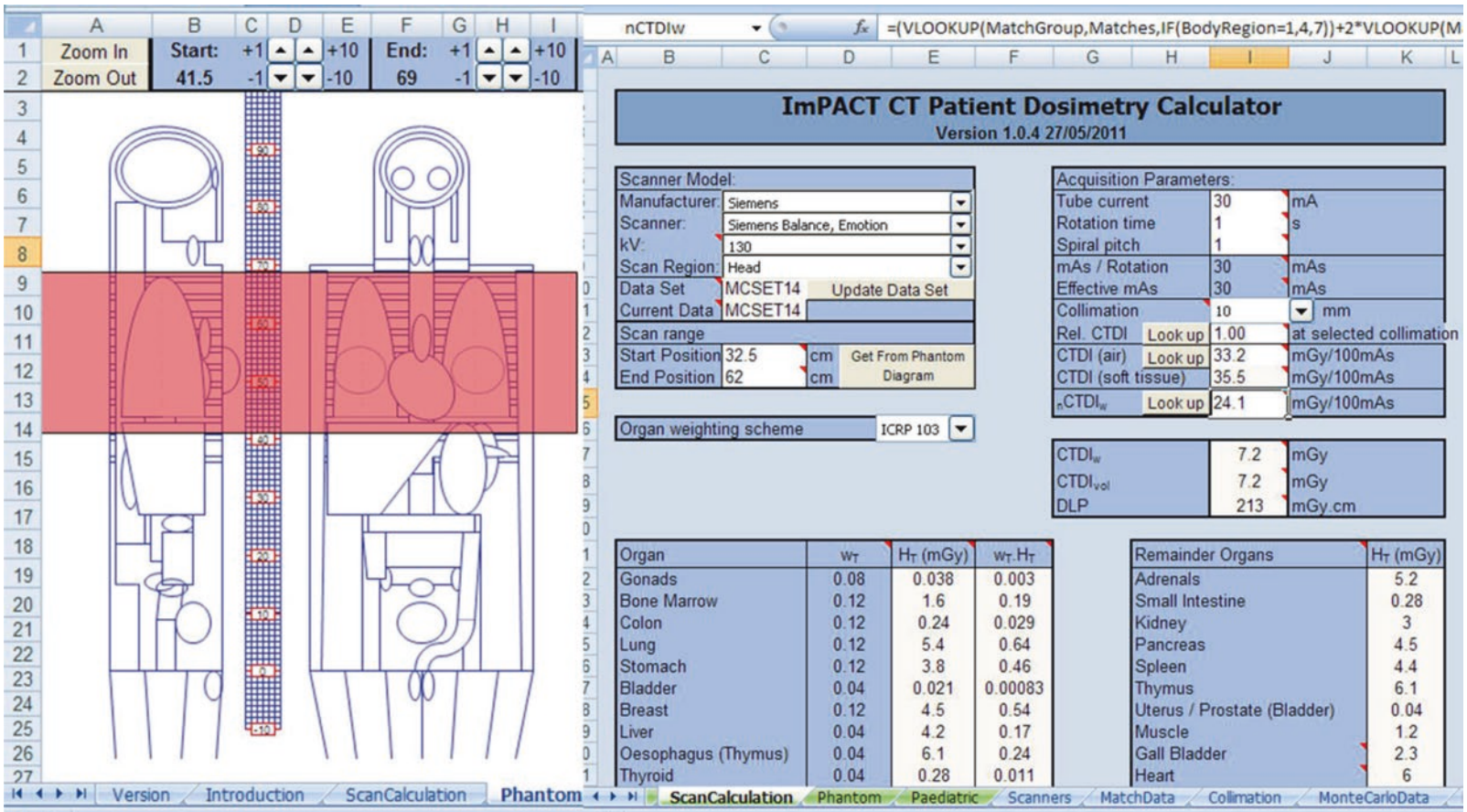


Fig. 2.30 The ImPACT Calculator

However, protocol review and optimisation is complex and requires a multidisciplinary approach as described in the AAPM guidelines [44].

2.4.6 Doses for Attenuation Scans and Whole-Body Scans in PET/ CT

Effective dose is the only way to compare the doses from the PET scan and the CT scan. The earlier sections demonstrated how much caution is required when comparing effective doses calculated using different methodologies in CT scanning. The radiation dose from the administration of the radiopharmaceutical is derived, for example, from ICRP 106 [45] or other sources as outlined in Chap. 3. This uses biokinetic models to generate coefficients of effective dose per MBq of administered activity (2×10^{-2} mSv/MBq), i.e. 8 mSv, for an administration of 400 MBq of F-18 FDG. Further caution must therefore be exercised when summing doses from PET and CT.

Furthermore, there are common descriptions of types of CT scan used in PET/CT. An attenuation scan is typically a low-dose, low mA scan. A diagnostic scan is typically described as a higher-dose scan which may also involve the use of contrast agents.

For whole-body PET/CT scans, doses of between 1.3 and 18.6 mSv are quoted for low-dose scans and diagnostic scans with contrast, respectively [46]. This results in a total examination dose of between 10 and 30 mSv. Effort and interest are going into minimising both CT dose and administered activity currently.

2.4.7 Quality Control of CT Scanners

As part of the life cycle of the equipment, safety checks, commissioning checks and routine quality control checks need to be carried out on the scanner. PET quality control checks are covered later in the book, but radiation safety and CT checks are included in the following section.

2.4.7.1 Commissioning Checks PET/CT

Checks should be carried out on the barriers installed within the facility. These can be assessed using a source of F-18 using the methodology described elsewhere [5]. Care needs to be taken in handling the source as a relatively high activity may be required to penetrate the barriers.

Further checks on safety signs and features should be made and recorded. A suggested list is shown in Table 2.8.

There is a full and rigorous testing programme for quality control that can be undertaken for all CT scanners (IPEM 91 [47], IPEM TGR 32 Part 3 [48], ImPACT Acceptance Testing [49], IAEA CT QA [50] and [51] IEC [52]).

Table 2.7 Scan parameters and impact on patient dose and image quality

Parameter	Impact on patient dose	Impact on image quality
kV	Increases with increasing kV ∞ kV^2 (approx.)	Noise decreases Contrast decreases
mA	Increases proportionally to mA	Noise decreases $\propto \sqrt{}_{mA}$
Rotation time (s)	Increases proportionally to time	Noise decreases $\propto 1/\sqrt{}_{sec}$
Pitch	Decreases as pitch increases (for constant mA)	Noise increases Depends on reconstruction algorithm and slice thickness
Beam collimation (beam width) (mm)	Decreases with increasing collimation	*Z*-axis resolution could decrease with wide beams for scanners with wider outer detector row elements
Detector collimation (mm) *z*-axis	Generally no effect	*Z*-axis resolution (image slice width) decreases with detector width
Scan length	Increases dose length product	No effect

The two most important measurements are dose and noise. The $CTDI_{air}$ is a valuable and quick test for quality control; in addition a full range of $CTDI_{vol}$ measurements for different scan conditions should be undertaken at acceptance and compared to the screen values. However, with the complexity of modern systems, a full test programme should be undertaken as listed in Table 2.9.

Table 2.8 Radiation safety checks for the PET/CT scanner and facility

Parameter
Room warning signs
Room warning lights: ready to emit radiation
Room warning lights: 'do not enter' prep/expose
Warning signals:
Mains on
Exposure light on control panel
Room protection:
General adequacy of protection
Adequate shielding of walls and doors
Surrounding dose rates meet design specification
Exposure switch
Labelling:
All controls clearly labelled
Focal spot position
Filtration/half value layer
Emergency off buttons
X-ray tube radiation leakage
Tube kilovoltage
Collimation x-ray to light beam and to detector

CT numbers are of interest to PET/CT scientists and should be included in any programme.

Conclusions

The design of PET/CT facilities has been described, considering some of the issues around working with PET radiopharmaceuticals that may be less familiar to those working in conventional nuclear medicine. The requirement for structural shielding around uptake bays and the scanner itself have been described, and a methodology outlined to calculate the thickness of barriers required to protect staff and members of the public within the facility and around it. The requirements to consider the external dose rate hazard, potential contamination and emergency situations are highlighted. The use of standard protection principles of distance shielding and time is emphasised and illustrated. Example calculations have been presented to help those unfamiliar with the methodology. The parameters required to perform the calculations have been summarised.

The potential for high staff dose is described, and approaches for assessment of whole-body dose finger dose have been discussed.

Fixtures, fittings and accessories within the facility have been explained with optimisation of radiation protection and minimisation of exposure and potential exposure described. Continued vigilance and review of staff dose have been advocated with some suggested

Table 2.9 CT quality control checks for the PET/CT scanner

Test	Comments	
CTDI in air	CT ion chamber/solid-state CT chamber	Compare with manufacturers specification
CTDI in phantom	CT ion chamber/solid-state CT chamber. IEC specified 16 cm and 32 cm phantoms	Calculate $CTDI_w$ Compare with specification
Image noise	Manufacturers phantom	Compare with manufacturers specification
Spatial resolution	Suitable phantom	Compare with manufacturers specification
CT number	Suitable phantom	0 for water −1000 for air Baseline for other materials
Imaged slice width	Suitable phantom	
Patient positioning vs. scan plane	Pins on surface of phantom	
Couch movement and position	Pins on surface of phantom	

methodologies to support keeping doses as low as reasonably achievable.

CT technology has been outlined in some detail to demonstrate the increase in speed and the number of detector rows between 1972 and the current day. The factors affecting dose in CT scanning – kV, mA and rotation time – have been illustrated. Image reconstruction, pitch, collimation and automatic exposure features have also been described. In order to fully understand each of these, a number of dose metrics have been defined and put into context against the technology and effective dose.

Diagnostic reference levels and optimisation of in CT have been discussed with the dose for attenuation scans and whole-body PET/CT scans.

The importance of quality control checks and references to support the checks required for CT have been supplied.

In conclusion, this chapter aims to supply sufficient information to support the safe design, operation and quality control of PET/CT scanners to those more familiar with conventional nuclear medicine.

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