Practical SPECT/CT in Nuclear Medicine

David Wyn Jones
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This book has taken over two years to produce and without the sustained hard work from the authors this would have not been possible. The authors have gone to great lengths to source and organise their information to deliver interesting, valuable and digestible narratives for the reader. On this basis we should like to thank all the authors. In particular we should like to express sincere gratitude to one of the authors - Professor Richard Lawson, particularly for his significant contribution to radiographer and nuclear medicine technologist education and training. Richard, a physicist from Central Manchester University Hospitals NHS Foundation Trust (UK), has freely given time and energy to encourage and develop others in the science of nuclear medicine. At the time this book was published Richard retired from the UK National Health Service. It is therefore fitting that Richard has contributed some excellent material within this book as this will serve as a valuable learning resource and clinical reference resource for years to come.

Foreword

Now that hybrid imaging is a well-established part of virtually all Nuclear Medicine departments, there is a great need for information on how to perform these studies to achieve optimal images. SPECT/CT is a fairly new addition to hybrid imaging and there is a paucity of information in one source, on how best to perform these studies. This book provides such a resource. This book was written to be a very practical guide to SPECT/CT imaging; consequently it is wide in scope. It provides information about both of the imaging techniques used in this hybrid system. There is detailed information about the theory and principles of imaging of gamma radiation, as well as imaging with X-rays. This is expanded to include the techniques of both SPECT and CT imaging and the imaging systems designed to acquire them in one sitting. This will be of enormous value for the vast majority of technologists and radiographers who have been primarily trained in one or the other of these modalities, but not in both. This is followed by well-illustrated clinical sections, giving examples of where this technique has been found to be useful. This includes specific recommendations of acquisition parameters, suggested injected activities and reconstruction techniques. There is also discussion of the radiation exposure resulting from these studies, including practical information on dose reduction techniques and acquisition parameters to help keep the radiation exposure a low as possible. This will be of enormous practical value as we work to keep the exposure to our patients as low as possible, while still maintaining high quality and clinical useful images. In short, this book will be a useful reference. It fills a gap in our current knowledge base in SPECT/CT imaging and should find a welcome place in every Nuclear Medicine department.

Chicago, USA — Gary L. Dillehay

Preface

Around the time of installing our first SPECT/CT system in North Wales with high-powered CT, it became apparent that there was very little published work available to assist those needing information on department design, procurement, installation, testing, commissioning, and clinical implementation. Consequently, this book was conceived in the context of providing a concise reference book that would bring practical and helpful information to radiographers and technologists who are relatively new to using SPECT/CT. Reflecting on the book contents, I feel it should also be of value to trainees in radiology, nuclear medicine, and medical physics and it may also have value to those already experienced in SPECT/CT.

Contributors to the book were drawn from a wide professional base, thereby reflecting the demands and scope of SPECT/CT. When selecting contributors, I was mindful of informing them of the potential readership so that the level and range was appropriate. I also brought some authors together in order for them to write chapters or subchapters collaboratively because I recognized that together they would bring more to the text than writing in isolation. I felt that having multiple perspectives on the same topic during the writing phase would assist in the creation of a richer and more accessible reference text. With this in mind, I drew together knowledge and expertise from colleagues in North Wales, the North West of England, the South West of England, the English Midlands, and Canada, but the core text should be applicable internationally. I hope that by bringing together these gifted individuals, the initial inspiration for a practical book for radiographers and technologists has been realized and that it shall have practical value in most nuclear medicine departments, and for associated professionals.

Wrexham, North Wales, UK — David Wyn Jones
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Abbreviations

D	Absorbed Dose
Al	Aluminum
A/C	Alternating Current
ADCs	Analog-to-Digital Converters
ALARA	As Low As Reasonably Achievable
μ	Attenuation Coefficient
AC	Attenuation Correction
ATCM	Automatic Tube Current Modulation
AEC	Automatic Exposure Control
ALMANAC	Axillary Lymphatic Mapping Against Nodal Axillary Clearance
Bq	Becquerel
BMI	Body Mass Index
E	Effective Dose
EMI	Electronic and Musical Instruments
EANM	European Association of Nuclear Medicine
CZT	Cadmium Zinc Telluride
^{11}C	Carbon-11
COR	Center of Rotation
CCTV	Closed-Circuit TeleVision
CT	Computed Tomography
CTAC	Computed Tomography Attenuation Correction
CIN	Contrast-induced Nephropathy
cps	Counts per second
CTA	Computed Tomography Angiography
CTDI	Computed Tomography Dose Index
CTF	Computed Tomography Fluoroscopy
CT#	Computed Tomography Number
CTPA	Computed Tomography Pulmonary Angiography
CCTS	Conventional Computed Tomography Scanning
DOH	Department of Health

DNA	DeoxyriboNucleic Acid
DICOM	Digital Imaging and Communications in Medicine
DVDs	Digital-Video-Disks
DC	Direct Current
DLP	Dose Length Product
ECG	Electro-Cardio-Gram
H	Equivalent Dose
FOV	Field-of-View
FBP	Filtered Back Projection
FD-CT	Flat Detector-Computed Tomography
FDG	FluDeoxyGlucose
^{18}F	Fluoride-18
FDA	Food and Drug Administration
^{153}Gd	Gadolinium-153
GEP-NETs	GastroEnteroPancreatic NeuroEndocrine Tumors
GI	GastroIntestinal
GINETs	GastroIntestinal-NeuroEndocrine Tumors
GE	General Electric
HCP	Health-Care Professional
HMPAO	HexaMethylProyleneAmine Oxime
HP-CT	High Power-Computed Tomography
HOCM	High Osmolar Contrast Media
HRCT	High Resolution Computed Tomography
HV	High-Voltage
HIS	Hospital Information Systems
HU	Hounsfield Units
HDP	Hydroxymethylene-DiphosPhonate
^{111}In	Indium-111
IAEA	International Atomic Energy Authority
ICRP	International Commission on Radiological Protection
IEC	International Electrotechnical Commission
IMRT	Intensity-Modulated RadioTherapy
ISD	InterScan Delay
IV	IntraVenous
^{131}I	Iodine-131
IR(ME)R2000	Ionising Radiation (Medical Exposure) Regulations 2000
IRR 99	Ionising Radiations Regulations 1999
IOCM	Iso-Osmolar Contrast Media
kVp	kiloVolt potential
Pb	Lead
LUTs	LookUp Tables
LEGP	Low Energy General Purpose
LEHR	Low Energy High Resolution
LEHS	Low Energy High Sensitivity
LOCM	Low Osmolar Contrast Media

LP-CT	Low Power-Computed Tomography
MRI	Magnetic Resonance Imaging
MSc NMI	Master of Science Degree in Nuclear Medicine Imaging
MAP	Maximum A Posteriori
MLEM	Maximum Likelihood Expectation-Maximization
MBq	MegaBecquerel
MIBG	Meta-Iodo-Benzyl-Guanidine
MIBI	Methoxy-IsoButyl-Isonitrile
MDP	Methylene-Diphos-Phonate
mA	milliAmpers
mAs	milliAmpere per second
mGy	milliGray
MTF	Modulation Transfer Function
^{99}Mo	Molybdenum-99
MPR	Multi-Planar Reconstruction
MSCT	Multi-Slice Computed Tomography
SVCT	Multi-Slice Virtual Computed Tomography
MPI	Myocardial Perfusion Imaging
NHS	National Health Service
NICE	National Institute for Clinical Excellence
NETs	NeuroEndocrine Tumors
NSAIDs	NonSteroidal Anti-Inflammatory Drugs
NCAT	NURBS-Based Cardiac-Torso
OJEU	Official Journal of the European Union
OSEM	Ordered Subsets Expectation-Maximization
PGD	Patient Group Direction
PACS	Picture Archiving and Communication System
PMMA	PolyMethyl MethAcrylate
PET	Positron Emission Tomography
PET/CT	Positron Emission Tomography/Computed Tomography
PET/MR	Positron Emission Tomography/Magnetic Resonance
POM	Prescription-Only Medicine
PIOPED	Prospective Investigation of Pulmonary Embolism Diagnosis
PSA	Prostate Specific Antigen
PUO	Pyrexia of Unknown Origin
QA	Quality Assurance
QC	Quality Control
RIS	Radiology Information Systems
RSNA	Radiological Society of North America
WR	Radiation Weighting Factor
ROI	Region of Interest
RCR	Royal College of Radiologists
SPR	Scan Projection Radiographs
SNR	Signal-to-Noise Ratio

SPECT/CT	Single Photon Emission Computed Tomography/Computed Tomography
SPECT/MR	Single Photon Emission Computed Tomography/Magnetic Resonance
SSCT	Single-Slice Computed Tomography
Sv	Sievert
SW	Slice Width
^{24}Na	Sodium-24
$Na^{99m}TcO_4$	Sodium Pertechnetate
SSTRs	SomatoSTatin Receptors
^{99m}Tc	Technetium-99m
^{123}Te	Tellurium-123
^{201}T1	Thallium-201
CsI(T1)	Thallium-activated Cesium Iodide
NaI(T1)	Thallium-activated Sodium Iodide
TLDs	ThermoLuminescent Dosimeters
3D	Three-Dimensional
W_T	Tissue Weighting Factors
2D	Two-Dimensional
USS	Ultra Sound Scan
^{235}U	Uranium-235
VIPoma	Vasoactive Intestinal Peptide tumors
V/Q	Ventilation Perfusion
VR	Virtual Reality
VRI	Virtual Reality Imaging
VDU	Visual Display Unit
WBC	White Blood Cells
WB	Whole Body
WL	Window Level
WW	Window Width
WHO	World Health Organization
^{123}Xe	Xenon-123

Part I
Scientific Principles

Chapter 1
Introduction

Peter Hogg

1.1 Introduction

This book is intended primarily for radiographers and nuclear medicine technologists who work with single photon emission computed tomography/computed tomography (SPECT/CT), but the book will be of value to other professionals within medical imaging. The intention of this book is to give a practical insight to SPECT/CT, to outline some of its uses, and also to explain how some of the more common procedures are undertaken. A fairly large amount of CT and CT-related material has been included because it is not uncommon for personnel working within nuclear medicine to have limited knowledge and experience of this modality. Purposefully we have not included a tremendous amount of information on nuclear medicine, so if the reader is novice to this field, they should consider reading a more general book about nuclear medicine alongside this text.

SPECT/CT evolved out of nuclear medicine – the addition of CT to SPECT allowed for important requirements to be offered to SPECT (attenuation correction). Serendipitously the addition of CT to SPECT also resulted in a large range of other uses for the hybrid SPECT/CT system. These additional uses did not become apparent for some years, and generally speaking, they came about through clinical and scientific staff trying out new ideas and new ways of working.

SPECT/CT evolved alongside other important advances within nuclear medicine and the broader medical imaging environment. In some respects, the progression of SPECT/CT technology and practice benefited from these advances; in other respects, it might be argued that some of these advances limited SPECT/CT's progress. Examples of advances that have likely facilitated the growth of SPECT/CT include the widespread introduction of position emission tomography (PET) and subsequently PET/CT; the broader provision of SPECT suitable radiopharmaceuticals, the

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further evolution of CT; the widespread use of picture archiving and communications systems (PACS); and the ever-increasing capacity of computers in conjunction with their ever-decreasing cost. It might be worth noting that the uptake of PET/CT, as opposed to PET without CT, came about because of the requirement for attenuation correction particularly for calculation of standard uptake values (SUVs). The technological advances made for PET/CT have no doubt advantaged the progression of SPECT/CT technology and perhaps influenced clinicians in their purchasing decisions made in relation to hybrid imaging more generally (i.e., SPECT/CT).

Today SPECT/CT has become standard in many imaging pathways, and in some circumstances, its use is written into national and international guidance. There is good reason for this; for instance, SPECT/CT can provide unique information that would be otherwise unavailable if SPECT and CT were used in isolation of each other. This unique information can assist in important decision making about patient diagnosis and management; consequently, the patient pathway and patient outcome can be altered positively.

The rationale for this book arose from discussions held with staff associated with the Diagnostic Imaging Research Programme and the Master of Science Degree in Nuclear Medicine Imaging (MSc NMI) at the University of Salford, Manchester, UK. The research program has two clinical research foci, one of which has a specific focus on an aspect of SPECT/CT, and within the MSc NMI, there is a module on hybrid imaging, which includes PET/CT and SPECT/CT. The authors of this book comprise of a professional mix that have scientific, technical, clinical, and/or theoretical knowledge of SPECT/CT. It is interesting to note that the clinical centers from which the authorship has been drawn represent the range of SPECT/CT technology which is currently on the market. This is important because this has allowed for that range of experience to be reflected into this book.

This book is divided into 5 parts and 13 chapters. Part I considers the physics and equipment which are necessary for the operation of a SPECT/CT system. Within this section, we have included information on X-rays; this addresses production and how they interact with matter. Attention is paid to X-ray production and the factors that can affect beam quality. This is important because in working practice the acquisition parameters can be manipulated in order to affect beam characteristics and ultimately CT image quality and radiation dose to the patient. As a reminder, an overview of gamma camera technology is given, and a particular emphasis is placed on the rotating camera/SPECT. As far as gamma camera construction and operation is concerned, within Chap. 3, the reader is pointed to further texts that would be necessary for a more detailed understanding. There is also a chapter dedicated to CT instrumentation and quality control. By the end of Part I, the reader should have a good grounding about the physics and technology of SPECT/CT. Part I was written by physicists and diagnostic radiographers. This professional blend was considered important, so the scope, level, and emphasis could be tailored to those professionals who would use the equipment clinically.

Part II considers practical information that should be taken into account when selecting and implementing a SPECT/CT system into a clinic. This section, written by physicists and diagnostic radiographers, draws on extensive experience from

those who have written business cases for and led the installation of SPECT/CT systems. Chapter 6 elaborates upon the factors that need to be considered when choosing a system to suit the clinical work to be undertaken and client group it serves. The demands of the client group in relation to SPECT/CT equipment on the market are also taken into account. Chapter 7 gives practical advice on facility/room design for housing a SPECT/CT system. Relevant regulations are taken into account along with a host of tips and hints.

Part III addresses the medical uses of SPECT/CT by drawing on clinical information and published research. This section was written by nuclear medicine physicians and radiologists who have considerable working knowledge of SPECT/CT. This chapter contains a large number of SPECT/CT cases and images. It brings to our attention a large collection of clinical uses of SPECT/CT from routine practice. We anticipate that radiologists and nuclear medicine physicians may find this chapter to be of interest.

Part IV considers a complex issue, that of radiation protection within the hybrid SPECT/CT environment; it also considers the optimization of CT acquisition parameters with a particular emphasis on image quality. This is an important chapter for nuclear medicine managers and personnel who have limited or no experience of CT. This section was written by radiographers and a physicist.

Part V outlines a practical perspective on using SPECT/CT. This section commences with a review of cross-sectional anatomy, with a particular emphasis on surface anatomy for CT positioning purposes. We have purposefully not included physiology because generally speaking this book is primarily aimed at a NM audience, and this knowledge has been assumed to exist. Chapter 12 considers X-ray contrast media and injection systems that can be used in CT. For contrast media, complications, clinical considerations, types, and contraindications are addressed. We have included these topics because we realize that contrast media is being used for some SPECT/CT investigations; equally we acknowledge that the CT component of diagnostic quality systems is being used for stand-alone CT examinations. Finally, Chap. 13 outlines a range of common SPECT/CT procedures that can be undertaken, with suggested protocols and hints/tips for getting a successful scan. This final chapter may be of particular value to clinical centers when setting up a new service or when wishing to make modifications to existing procedures. This section was written by radiographers and nuclear medicine technologists.

1.2 Origins and Evolution of SPECT/CT

SPECT/CT comprises of two separate technologies combined into one physical unit in order to produce CT and nuclear medicine data in the same session. SPECT/CT is one of several hybrid systems currently on the market; of the hybrid installations that have already taken place, the most prevalent is PET/CT. It is worth noting that PET/MR is now commercially available, and a limited number of these systems have been installed. SPECT/MR has also been proposed, and it might be interesting

to note that optical/PET and optical/SPECT are also being considered. Optical imaging is a relatively new technique which can image physiology. Triple modality hybrid systems are also being discussed.

SPECT/CT combines functional (SPECT) and anatomical (CT) imaging; the added value of CT is that a bespoke attenuation map can be generated and applied to SPECT image data to account for lost counts through self-attenuation. Aside attenuation correction, to date, another purpose of SPECT/CT has been to combine physiology and anatomy into a single registered image. Registered images are said to be of value in certain diagnostic situations; additionally, they appear to have a growing value within planning for radiotherapy. Recently, discussions within the medical imaging community have considered whether there might be value in combining two physiological imaging systems and similarly combining two anatomical imaging systems [1]. Publications are starting to emerge to speculate the value of PET and SPECT combined with optical imaging. It is worth noting that the integration of MR with PET or SPECT has been slow to move forward, and various factors have accounted for this. For instance, MR is unable to provide data for attenuation correction, unlike CT; also there continues to be inherent problems of enabling MR to work easily alongside PET or SPECT, and vice versa. Each machine has the potential to interfere with the successful running of the other.

Although the concept of SPECT/CT was reported over two decades ago [2], it is only in recent years that these hybrid systems have gained widespread popularity and use in the clinical routine. The initial intention for having an integral SPECT/CT system was for attenuation correction of SPECT data; in this respect, the CT unit represented an important advancement from the radioisotope and/or mathematically based alternatives. Gd-153 is an example of a radioisotope external beam method for empirically deriving a bespoke attenuation map for SPECT data. Limitations of the radionuclide approach surrounded the low radiation output when the Gd-153 became older (through physical decay) and also the cost of replacing the Gd-153 sources. An alternative to the radionuclide approach was mathematical modeling. Modeling made assumptions about body composition, and when the assumptions were not met the attenuation correction data could be highly inaccurate. For instance, one approach assumed that all pixel values would be the same and when values varied significantly large attenuation correction errors would be incurred. One body area where this approach had limitations was the chest; by contrast, the brain is fairly homogenous and less prone to such errors.

Moving to the hybrid SPECT/CT environment comprised of two major stages, the first did not involve the physical fusing of the two technologies while the second did. In the first stage, image data from CT and SPECT were acquired on machines that were situated in physically different locations. The two datasets were then brought together and registered to enable CT attenuation maps to be applied to SPECT and/or images to be fused for display. This approach was fraught with problems, including difficulties in aligning datasets from different machines and the potential for poor scheduling between the two scans – in some instances, the SPECT and CT images could be generated days or even weeks apart. Such time differences could render the datasets difficult to compare, not least because the pathology could

have progressed and/or spread; consequently, the CT and SPECT data would be reflecting a real difference that could not be correlated/registered.

The first commercially produced SPECT/CT system was introduced by General Electric in 1999. The CT component comprised of low-dose fixed output with a maximum tube current of 2.5 mA. In some respects, this initial commercial response to the SPECT/CT market was relatively easy to introduce into the clinical routine. Aside the complexities of X-ray radiation protection, the operation of the CT component was quite easy, as it had preset values with one intention in mind (attenuation correction). As experience with fixed output systems grew, clinical staff experimented with the low-quality CT images that were produced, and it started to become apparent that these images may have clinical value. In the first instance, their value surrounded assisting with the anatomical location of hot or cold areas seen on SPECT. However, the limitations of the fixed output CT machine's image quality meant that its clinical value was restricted.

Since the early days of limited image quality, the CT component of the SPECT hybrid system has evolved considerably, and today systems offer multislice potential of equal ability to stand-alone diagnostic CT machines. Sitting between the high-end diagnostic quality and fixed output CT systems are a number of midrange machines. These midrange CT machines offer limited acquisition flexibility. For instance, there may be the option to select from a limited range of mA and kVp values; these parameters can affect image quality. Figure 1.1 illustrates three CT slices taken of the same anthropomorphic chest phantom. The displayed slices are at approximately the same level. Image A is from a fixed output single slice CT machine, image B is from a midrange 4 slice CT machine (using 2.5 mA), and image C is from a diagnostic quality CT machine (16 slice). On comparing the three slices, it is clear that resolution differences are evident. It is worth noting that the anthropomorphic phantom is stationary during CT image formation, in a human because of long exposure times, there could also be blurring due to breathing and heart motion.

In the early years of SPECT/CT, debates were held on whether this hybrid technology might be a passing fashion simply because its initial intention was limited to the attenuation correction of SPECT data and that application was restricted to certain procedures and often on specific patient populations therein. Some considered SPECT/CT to be a very expensive solution to attenuation correction, assuming that this was its sole purpose. Not surprisingly, questions were raised about value for money. However, the move toward midrange and diagnostic quality CT scanners opened up new possibilities which were capitalized on within business cases and working practice. For instance, the high-end diagnostic quality CT scanner could be operated as stand-alone; consequently, if required, it could provide an alternative service to the main X-ray department CT scanner. This would have value for imaging patients when the main CT unit was unavailable (e.g., during servicing and breakdown); also it could provide additional CT capacity when required. It may be worth noting that the provision of additional capacity has been adopted by some centers, as they now run CT sessions in certain evenings of the week after the radioisotope scanning has finished for the day. Business cases have been made for

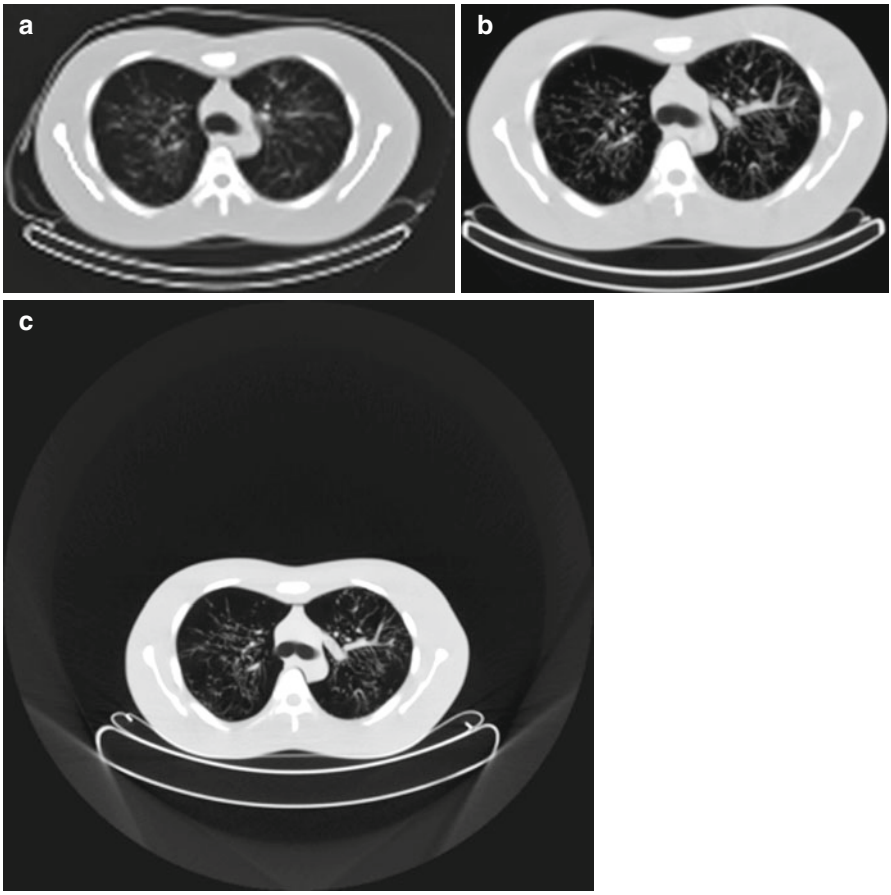

Fig. 1.1 (**a**) Low-resolution/fixed output. (**b**) Medium resolution. (**c**) High-resolution/diagnostic quality

SPECT/CT systems on this basis, and for a radiology department, it might be more cost-effective to install a high-end diagnostic quality SPECT/CT system than purchase an additional stand-alone CT system and a lower-end SPECT/CT system as well. This decision would of course be influenced by CT workflow.

Having integral diagnostic quality CT with SPECT has resulted in new possibilities for imaging and patient workflow. For instance, diagnosis of some conditions which previously may have required multiple patient visits for imaging using the different modalities can now be consolidated into one hospital visit. This has particular benefits to patients in that one hospital visit, rather than two or three, would be required. This approach reduces travel costs to the patient, and if they were quite unwell, then the stress and effort of having many hospital visits would be reduced too. Depending upon the patient and their condition, this could mean hospital savings too, for instance, ambulance transport may be required on one rather than

several occasions. However, this condensed approach has brought new patient related problems which do require consideration, for instance, the patient may find out much more quickly what is wrong with them where previously it could take several days for the information to be revealed, thereby allowing the patient a chance to acclimatize to any unwelcome and stress-inducing news.

Until the introduction of SPECT/CT, the only alternative way into the hybrid market for a radiology department was PET/CT. PET/CT scanners remain more expensive than SPECT/CT systems, this being especially true when the cost of the ancillary equipment and consumables (e.g., cyclotron and the cost of the PET radiopharmaceutical) are taken into account. Being cheaper, SPECT/CT allowed for a radiology department to enter the hybrid imaging market at a fraction of the price of PET/CT. Today, when replacing a gamma camera, it is common that consideration is given to whether a SPECT/CT system should be installed as the gamma camera replacement. Various factors come to play when making this decision, including the available finance and the clinical work that will be conducted on it. SPECT/CT has particular values, as will be highlighted in this book, and the case for purchasing a system should take these into account. Of course replacing a gamma camera with a SPECT/CT system comes with hidden costs, and these should not be overlooked. For instance, there will be a requirement to install secondary radiation shielding into the room walls and doors. Also there may be a need to provide a shielded area within the imaging room for the imaging staff, or more commonly provide a shielded control room from which the patient can be observed through lead glass and/or CCTV.

Unlike PET/CT, SPECT/CT had a surprising start to its emergence into the clinical routine. By early 2008, in spite of there being a growing number of SPECT/CT installations, the literature was highly deficient with fewer than 150 journal publications being available worldwide. Of these, many were speculative; empirical pieces did exist, but many had very small patient numbers. Around that time, the notion of evidence-based practice for SPECT/CT was clearly not possible, which on reflection is quite concerning. This was recognized as a problem within the European Association of Nuclear Medicine conference in 2008. Various explanations have been put forward since 2008 about the literary deficiency around that time with one of the more plausible being that eminent national and international scientists had focused their energies into PET/CT, rather than SPECT/CT, perhaps because of PET/CT's acknowledged value in many aspects of current and future medical practices. Today the picture is brighter because much more SPECT/CT literature has been published into journals, and a growing number of books/book chapters dedicated to SPECT/CT have started to appear. It might be worthwhile remembering that quality literature is important to the evolution of medical techniques and medical devices, and this obviously includes SPECT/CT. Health-care systems in various countries have adopted the philosophy of evidence-based practice. Ideally evidence should be generated prior to the widespread introduction of a new technique, as this protects the patient from harm and inappropriate practice. Also it protects the employer and health-care team from clinical negligence claims which can arise from inappropriate care and management.

1.3 The Virtues and Challenges of Change

Depending upon how imaging services are resourced and managed, it is possible that the fusion of SPECT and CT would bring together different cultures and different disciplines. For instance, nuclear medicine (SPECT) personnel may have to integrate and work closely with radiology (CT) personnel. Philosophically speaking, this union can bring tremendous opportunity. For instance, the sharing of different perspectives, experience, and ideas can present a powerful catalyst for innovation and advancement; indeed such innovation and advancement are evident in practice and in the SPECT/CT literature.

On a practical level, the fusion of different cultures and practices can present challenges, and these need to be acknowledged and, as required, resolved in an amicable fashion. For example, radiographers and radiologists who practice nuclear medicine are likely to be able to perform/interpret CT examinations; for nuclear medicine technologists and nuclear medicine physicians, they may need to develop CT skills. Similarly, the technical teams that support the equipment, including physicists and technicians, may also have specific backgrounds and therefore be specialist in only one of the two technological areas. In itself, the knowledge and skill differences must be seen as relatively minor problems as appropriate education and training can be sought to redress any perceived deficiencies. Admittedly there may be conflicts in role definition between professional groups which will likely require clarification and resolution. Such political negotiations can be time consuming and stressful as they will no doubt take into account emotional responses which have good grounds from a personal perspective, but they do not necessarily have good grounds from a service delivery and patient standpoint.

There is a very good example of two professional groups coming together to agree practice standards in hybrid imaging. While the field is PET/CT, the principle would be the same as SPECT/CT. In 2004, with an emphasis on PET/CT, America addressed a competence and practice imbalance. The American Society of Radiologic Technologists together with Society of Nuclear Medicine Technologist Section produced valuable documentation on the knowledge and skills required to operate a hybrid system, to include CT [3]. The proposal was that multiple pathways should be created to train professionals, the multiple pathways being necessary because of the different routes people may take into hybrid imaging (e.g., from nuclear medicine or from radiography/radiology). Subsequent to this scientific and professional bodies in other countries have responded similarly with proposals on what skills and knowledge may be required for working in a hybrid environment [4].

Training/competence is only part of the solution to overcoming change challenges. In some countries, regulatory (legal) restrictions on "who can do what" may inhibit nuclear medicine technologists and others from making CT exposures. Considering this matter from a patient-centered stance, the emphasis should be on competence to practice, because it should be a matter of patient safety and well-being rather than a professionally focused argument which asserts that only certain professional groups can do the task/assume the responsibility. Expanding the

patient-centered argument a little and illustrating through a practical example, if we entrenched our thinking to only a small group of professionals being able to make the CT exposure, then instances would arise when the patient would be disadvantaged. For example, the nuclear medicine technologist performs the nuclear medicine SPECT study, and then they need to ask a radiographer to make the CT exposure. Clearly this is a poor use of radiographer time, and it underestimates the potential of the nuclear medicine technologist. Additionally, the possibility of delays can arise to the patient journey, and the additional cost of having two instead of one professional involved would add to the cost of the procedure. Where regulatory limitations exist, thorough consideration should be given to reaching sensible solutions. Such solutions would certainly include making proposals on "who can do what," legally speaking, in order to provide safe, equitable, and appropriate service to patients. Being pragmatic, perhaps this could be addressed on a sliding scale, on which low-dose fixed output CT for AC would require less extensive training, and by contrast, high-end diagnostic quality CT would require considerably more extensive training.

The integration of two quite different imaging modalities, such as SPECT and CT, brings together different cultures and disciplines which can have quite dissimilar clinical perspectives. For instance, some years ago, PET in isolation of CT showed great promise in the diagnosis of certain cancers. For cancer management, because of clinical experience and research, evolution of our understanding in cancer management made us realize that PET should not be conducted without CT. Initially this line of thought was for attenuation correction purposes, but increasingly the use of diagnostic quality CT with PET is becoming an essential requirement too. Ultimately PET (with CTAC) and diagnostic CT images started to be interpreted together within the same reporting session. This meant that those with a traditional interest in physiological imaging started to work closely with those that had a traditional interest in structural appearances. Not surprisingly when physiological and structural interpretations began to be synthesized in the same physical space, questions between professionals started to arise on how the information from the combined study might be best explained and then portrayed. In fact, the debate has started to go further than this, for instance, the growing integration of hybrid imaging and radiotherapy has meant that the clinical images (physiological and structural) and associated data have started to play an increasing role in radiotherapy planning. For cancer treatment planning, interpretation and application of the hybrid images have now started to include members of the oncology/radiotherapy team. While the points raised here are likely to be more pertinent to PET/CT rather than SPECT/CT, it does give us clues on "what next" for SPECT/CT and the challenges which are likely to lie in its future.

Aside regulatory, cultural, and competence matters, important changes to working practice are likely to be imposed through the introduction of CT, and this could have a significant impact onto the patient experience. For SPECT it has been common practice for personnel to remain inside the imaging room throughout the examination. When working practice is optimized in accordance with regulation and its guidance, this practice incurs negligible risk to personnel. From a patient standpoint, having personnel within the imaging room can bring benefits to the procedure

itself and also add to the well-being of the patient. Practical well-being benefits may arise from the close proximity of staff in cases when the patient needs immediate physical help and/or emotional support.

Well-being can be related to a number of factors. For instance, it might be that the patient does not feel isolated because they would know that personnel are in close physical proximity and also within easy earshot. The introduction of CT, particularly for high-end and therefore higher-dose diagnostic quality systems, will bring with it the need for a personnel control room. This control room is normally separate to the imaging room. This leads to temporary separation of personnel and patient. In this circumstance, to counterbalance patient care/well-being requirements, there may be a need to adopt different working practices to ensure patient needs are identified, met, and monitored in a fashion that limits X-ray radiation risk to staff yet supports the patient adequately in terms of their physical and emotional needs. Clearly the transition from SPECT to SPECT/CT will require radiation risk to staff to be balanced against patient care and well-being demands. Suitable working arrangements will need clarifying and reflecting into practice.

Remaining with radiation protection for a moment let us think about some basic considerations about patient radiation dose. The addition of CT to SPECT incurs additional radiation to the patient, and the medical practitioner requesting and/or being responsible for the imaging procedure must ensure the additional radiation is justified. Optimization of (CT) dose will also be important. It has been reported in the literature, and the popular press, that radiation exposure for medical reasons has been on the increase and a great contributor to that growth has arisen from CT. The radiation burden associated with CT can be significant, and this is particularly true for high-end diagnostic quality CT systems. In some circumstances, the requirement for CT as part of the SPECT procedure can be written into formal documents for radiographers and nuclear medicine technologists to refer to. Such documents can explain when it must be used. Equally there will be circumstances when CT would not be justified, and these situations should be clarified. Clarification appears to be adequate in the instances where attenuation correction maps are required. However, there are times when further imaging may be required subsequent to SPECT and the decision to take the CT image should be patient dependent. This decision would be taken in light of the clinical question, the clinical background, investigative tests undertaken already, and the possible meanings of the SPECT information. Such decisions need consideration prior to implementing SPECT/CT into routine practice, and in particular there would be a need to determine how those decisions will taken and by whom.

With CT imaging in mind, consideration should also be given to alternative diagnostic imaging approaches, such as plain X-ray. While it may be convenient to use the CT component to acquire a structural image, it is possible that a plain X-ray image may meet the clinical need at a faction of the radiation dose. Additionally, when considering plain X-ray versus an image produced by *low-resolution* CT (intended for CTAC), it could be that better clinical information would be gained from the plain X-ray. Examining the literature, it is evident that inappropriate practice has occurred, whereby poor quality CT images have been produced, and these

have had to be followed up immediately by plain X-ray imaging so that a fuller diagnosis can be reached.

Nuclear medicine procedures have a long history and culture of basing the administered radiation dose upon published information. In some countries, national guidance [5] is available to direct medical practitioners, and others, on the amount of radioactivity which should be administered for each examination. These amounts have tended to be derived through research; consequently there is an evidence-based approach to dose limitation/optimization in nuclear medicine. The guidance also proposes ways in which the administered radioactive dose might be varied and the circumstances where such variations could be clinically justified. For instance, compared with an adult, the quantity of dose would be reduced for children, and the reduction would bear relationship to the child's physical makeup (e.g., weight). Latitude for the medical practitioner would be facilitated, in that they might decide to increase or reduce the dose depending upon the weight of an adult – an obese patient may be administered more dose, for instance. Additional dose guidance is also available in terms of the technical demands of a nuclear medicine procedure, for instance, a study involving SPECT may involve administering a higher dose than for a planar study.

X-ray imaging has a different history and culture, and while the principles of dose limitation and optimization are equally important, the principles upon how the dose reference levels are achieved can be quite different. For instance, within the UK, X-ray dose reference levels are based upon surveys of current practice, and this applies to CT. Within SPECT/CT, the optimization of dose is more complicated because there are at least two reasons for performing the CT component – to attenuate correct SPECT data and/or to produce an image that will have diagnostic value. It is likely that the dose required for AC will be lower than for imaging; however, the literature is not clear on this matter, and more research needs conducting.

The ultimate decision on the purpose of the CT acquisition would be medically based, and the amount of radiation incurred would bear direct relation to that decision. For instance, the use of CT for attenuation correction alone would likely incur a reduced dose when compared to using CT for creating an image of diagnostic quality. If the decision surrounds only attenuation correction, then one set of acquisition factors may be used. However, if there is a need for a CT image of diagnostic value to be obtained as well as data for attenuation correction, then a higher amount of radiation will likely be needed. This should be reflected into the acquisition parameters/protocol.

An area of interesting practice that has started to develop in some centers involves the formal review of the lower-quality/lower-dose CTAC image for detecting incidental pathology. This practice is likely to be of particular value for the CTAC image of the lung fields because of the inherent high contrast of that body part.

When using the CT beam to produce an image (with or without attenuation correction), there will be a need to conduct optimization experiments to ensure that the image is of a standard fit for purpose. This is a complex task as the optimization process cannot rely simply on physical measurements (e.g., image noise levels using physics phantoms). Various methodologies have been proposed in the literature to assist in the optimization of image quality and radiation dose, and a robust approach

must be taken which takes into account valid and reliable visual perception measures as well as the physics measures. Obviously this optimization work should be completed prior to clinical use.

1.4 Future Potential of SPECT/CT

So far we have considered the origins and evolution of SPECT/CT; now we shall take a look into its future. Historical data combined with contemporary information can be used to predict the future, and this approach has long been used in the business sector; it is known as trend analysis. In some respects, trend analysis can be self-limiting as it can restrict creative thought by tending to focus our thoughts onto successes and failures [6]; therefore, it can be argued that trend analysis has a predisposition to point us toward what might be considered as fairly obvious conclusions. Consequently, using trend analysis, we can speculate that SPECT/CT will continue to have clinical value for attenuation correction and that registration and display of images will continue to have importance for precise lesion location and improved diagnostic accuracy.

A more speculative approach to predicting the future of SPECT/CT is taken here, and an emphasis is placed on one potential application which is starting to emerge in the literature – molecular imaging [7]. Molecular imaging has the potential for a broad application range, and it can be used in all aspects of medical imaging – from screening through to diagnosis. It can also be used to measure the response to therapy. For certain cancers, current approaches tend to be based on CT or MRI data to measure changes in tumor size. This approach has limitations in that relying on size response could take weeks or longer for the impact of therapy to be realized. Consequently, molecular imaging has the potential to improve therapeutic response evaluation by providing an indication of that response much earlier in time and well before anatomical changes become perceptible. This is one reason why PET/CT is growing in importance for radiotherapy planning [8].

Molecular imaging involves the use of molecular probes to target and demonstrate biological processes, including pathology. Probe creation involves identifying a specific feature of the target molecule that characterizes the disease to be imaged. The process of developing a probe is time consuming and costly. Within nuclear medicine, examples of these probes include radiolabeled antibodies, ligands, and substrates. Two main approaches exist in molecular imaging – direct and indirect imaging. Direct imaging involves targeting the molecule of interest; indirect imaging concerns the targeting of gene products. Probes for direct imaging interact with the target, and four criteria are essential for this to occur: first, the availability of high affinity probes; second, the ability of these probes to overcome biologic delivery barriers; third, the availability of amplification strategies; and finally, the availability of suitable imaging techniques.

Scientists and clinicians argue that we are at the start of a paradigm shift – we are heading toward personalized medicine. The underlying assumption for this shift is

that the present generalized approach of *one solution fits many* is not the way things should be done and molecular imaging (and therapy) is in the forefront of this transformation, especially for cancer management. The main advantage of molecular imaging is its ability to characterize diseased tissues without biopsies or surgery; with this information, a more personalized treatment plan can be derived. Alongside this, current health-care demands are focusing on the need for better approaches to early disease detection and molecular imaging excels in this.

From humble beginnings, molecular imaging has rapidly gained popularity and not just within a research context. Molecular imaging is concerned with the visualization, characterization, and quantification of biological processes in living tissues at the subcellular and cellular levels. Molecular imaging is so specific that particular DNA can be targeted and then imaged. From this, information can be obtained about diseases within their native environment. This is contrary to historical and contemporary approaches which tend to involve the removal and examination of diseased tissue within a foreign environment (e.g., test tube) and/or the in vivo imaging of diseased tissue using relatively simple techniques which demonstrate changes to anatomy; in some instances, such approaches can only image disease if fairly gross changes have occurred. Molecular imaging is different, in that it has the potential to demonstrate disease well in advance of the traditional approaches – consequently, it permits much earlier detection of disease. It may therefore help to identify a disease long before signs and symptoms appear. Speculation suggests that molecular imaging will have a major economic impact due to earlier and more precise diagnosis.

Various approaches to molecular imaging can be taken; presently, the most common is PET/CT. Other technologies are capable of molecular imaging too, including MR, ultrasound, optical, and, of course, SPECT/CT. Table 1.1 indicates a number imaging modalities that can be used for molecular imaging along with some of their advantages and limitations. Table 1.1 indicates that SPECT/CT has many molecular imaging advantages and its limitations are not overly extensive.

In medicine, there is an increasing tendency to employ the principles of translational research in order to speed up the process of transferring findings of research into practice; in the context of molecular imaging, PET/CT has excellent translational research potential, and SPECT/CT has good potential for this too. This is an important observation, for without good translational research potential, the uptake of molecular imaging, utilizing SPECT/CT, into the clinical routine would be severely limited. It is said that this is likely to be the case because PET and SPECT are the most generalizable of all the available molecular imaging approaches. Therefore, as molecular imaging continues to advance, we should see PET/CT and SPECT/CT remaining to be a central part of the molecular imaging revolution. Aside the high potential for translational research, SPECT and PET have a minimal potential for adverse biological effects, adding further to their value in routine practice. While it is known that there is an extensive and growing list of PET radiopharmaceuticals for molecular imaging, it should be noted that there is also a good and growing range of SPECT radiopharmaceuticals available suited to molecular imaging too; some examples are given in Table 1.2 [9].

Table 1.1 Imaging modalities that can be used for molecular imaging along with some of their advantages and limitations

Modality	Advantages	Limitations
PET/CT [10]	Quantitative	Ancillary equipment is required (e.g., cyclotron)
	High sensitivity	High cost
	Isotopes can proxy for naturally occurring atoms	Low spatial resolution
	Possibility of whole body imaging	Patient irradiation
	Excellent translational research potential	Long image acquisition times
	Very good range of molecular probes	
SPECT/CT	Good range of molecular probes	Low spatial resolution
	Possibility of whole body imaging	Patient irradiation
	Good potential for translational research	Long image acquisition times
Optical [10], fluorescence, and bioluminescence	Highest sensitivity	Whole body imaging is not possible
	Quick and easy	Low spatial resolution
	Low cost	Limited depth penetration
	No ionizing radiation	2-D imaging only
	Relatively high throughput	Relatively surface weighted
		Limited potential for translational research
MR [10]	Very high spatial resolution	Relatively low sensitivity
	Combines structural and functional imaging	Long scan and post-processing time
		High cost
Ultrasound [10]	Real time	Limited spatial resolution
	Low cost	Whole body imaging not possible
	No ionizing radiation	Ultrasound scanning can be highly operator dependent
	Can be used internally and externally	

Molecular imaging is becoming an invaluable element of the preclinical evaluation of X-ray contrast agents and the development of new drugs [10]. Progress in molecular imaging has allowed for the use of noninvasive (imaging) techniques into drug discovery and development. Drug development concerns the bringing of a new drug to clinical practice, and it comprises of a number of stages, including preclinical work and trials on humans. The cost of developing a drug for clinical practice can be immense, and only a small percentage of the compounds identified in the initial stages of development reach the clinic. The process is also lengthy. Prior to trials on humans, data has to be analyzed to establish safety, toxicity, pharmacokinetics, and metabolism. As part of the preclinical process, the new compound will be analyzed for its suitability to be made into various forms (e.g., tablets, aerosol, and injectable forms). Drug development has to satisfy legal requirements in order to gain a license to be used on humans. To achieve this, many tests have to be conducted to understand its toxicity; this includes determining the effects on major

Table 1.2 SPECT radiopharmaceuticals suited to molecular imaging

	Molecular target	Radiopharmaceutical	Clinical use
Oncology	Carcinoembryonic antigen	^{111}In altumomabpentetate	Colon cancer
	Tumor-associated glycoprotein	^{111}In satumomab pendetide/ OncoScint	Colorectal or ovarian cancer metastasis
Cardiovascular	Perfusion	^{99m}Tc teboroxime/ Cardiotec	Myocardial perfusion
	Perfusion	^{99m}Tc tetrofosmin/ Myoview	Myocardial perfusion
Neurological	Amphetamine receptor	^{123}I-iodoamphetamine	Neurodegenerative disorders; regional cerebral blood flow
	Phosphatidylserine	^{99m}Tc-HYNICeannexinV	Dementia
Musculoskeletal imaging	Hydroxyapatite crystals	^{99m}Tc oxidronate(HDP)/ Osteoscan _ HDP	Bone
Apoptosis and/or cytotoxicity		^{99m}Tc-annexin VMRI	
Prostate-specific membrane antigen		^{123}I-labeled glutamate-urea-lysine analogues;161 ^{99m}Tc-chelates162	
Estrogen receptor		^{131}I-tamoxifen; tridentate ^{99m}Tc(I)-estradiol-pyridin-2--yl hydrazine	

organs such as the heart, lungs, and brain. Many of these tests can be made within test tubes (in vitro); however, a number of the tests have to be performed on living and in vivo tissue. The latter would involve the use of experimental animals and humans, and molecular imaging has a role to play here, for example:

- Confirmation that a drug has reached its target
- Determining drug pharmacokinetics
- Determining drug biodistribution
- Assessment of drug–target interaction

Molecular imaging can provide data to support proof-of-principle and proof-of-concept analyses. Proof of concept relates to the drug's positive effects; this would likely translate into beneficial clinical values. Microdosing involves the initial experimental work conducted on humans, in vivo. The principle behind microdosing is that the quantity of drug administered to the human would be so small that volunteer safety is not comprised. However, the microdose would be sufficient so as to provide information about in vivo drug behavior. Molecular imaging, particularly using PET/CT and SPECT/CT, has an important and growing role in microdose work; this is because the small quantities of administered drug can be assessed through these highly sensitive imaging techniques. Microdosing studies can therefore be used to

assess the pharmacokinetic profile of a drug using imaging techniques. To achieve this, the drug has to be radiolabeled and then administered. Dynamic in vivo pharmacokinetics of the drug can then be gathered and analyzed. On the basis of this investigation, a decision can be taken on whether to precede to a phase I clinical trial – this is the first stage of testing on humans. Limitations of radionuclide approaches in microdose studies surround the short half-lives; in some instances, in vivo pharmacokinetic data needs to be gathered for lengthy periods that go well beyond the potential of PET or SPECT imaging.

SPECT/CT can provide a way to attain information about the in vivo properties and therapeutic efficacy of new drugs. SPECT/CT may therefore assist in bringing new drugs to the market place more quickly, more cheaply, and more safely.

1.5 Conclusion

This initial chapter has set the scene for the book. From a practical standpoint, it has given an overview of the contents of the book and also given a basic insight into how SPECT/CT came about and where it is today. This chapter has also considered the future of SPECT/CT. While the future of SPECT/CT could be viewed as somewhat predictable, by standing back and taking a more speculative and visionary approach, it is possible that SPECT/CT could play a major role in the delivery of molecular and personalized medicine.

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Chapter 2
Basic Discussion on Electromagnetic Radiation: Gamma Radiation in Relation to SPECT Imaging

Richard S. Lawson

2.1 The Atom

All materials are made up of *atoms*, which at one time were thought to be the smallest particles of matter (the name comes from the Greek *atomos* meaning indivisible), but now we know that the atom actually comprises a cloud of *electrons* circulating around a nucleus at its center. The nucleus is made up of *protons*, which have a positive electric charge, and *neutrons*, which have no electric charge; see Table 2.1. Since the electrons have a negative charge, they are held in their orbits by electromagnetic attraction from the protons, and in a normal atom the number of protons and electrons is equal so that the charges balance out and the atom is neutral overall. Atoms are very small but the nucleus is much smaller still, so the atom is actually mostly empty space. (If an atom was magnified to the size of a balloon, the nucleus would still only be as big as a speck of dust inside it.) However, the electrons are very light compared with the protons and neutrons, so almost all of the mass of the atom is concentrated in the nucleus. The protons and neutrons in the nucleus are held together by strong nuclear forces which are sufficient to overcome the repulsive electromagnetic forces that tend to push the positively charged protons apart.

2.1.1 Electron Energy Levels

The electrons in an atom can only exist in certain energy levels or shells, each of which has a definite energy, and each energy level can only hold a specific number of electrons; see Fig. 2.1. Normally electrons fill the available energy levels starting

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Table 2.1 Properties of some fundamental particles

Name	Mass (MeV)	Mass number	Charge
Proton	938	1	+1
Neutron	939	1	0
Electron	0.5	0	−1
Positron	0.5	0	+1
Neutrino	Negligible	0	0
Photon	0	0	0

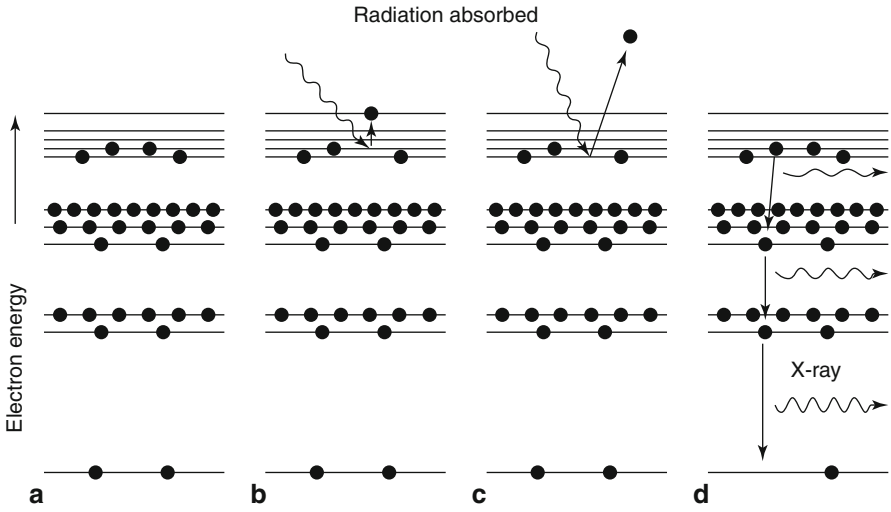

Fig. 2.1 Electron energy levels in an atom. (**a**) The ground state – all the lowest levels are filled. (**b**) Excitation – absorbed radiation results in an electron being raised to a higher level. (**c**) Ionization – absorbed radiation results in an electron being ejected from the atom. (**d**) Emission – electrons cascade down to fill a vacancy at a low level and energy is emitted as X-rays

from the lowest, and so in its lowest energy, or ground state, the atom has all its lowest energy levels filled (Fig. 2.1a). However, if an atom absorbs some energy, one of its electrons can be raised to a higher energy level, known as *excitation* (Fig. 2.1b), or ejected from the atom completely, *ionization* (Fig. 2.1c). Following either of these processes, a vacancy can be created in one of the energy levels. The atom can then regain a stable configuration by letting an electron from a higher energy level fall down to fill the vacancy, and in doing so it will get rid of the energy difference by emitting a *characteristic X-ray* (Fig. 2.1d).

2.1.2 Nuclear Energy Levels

In a similar manner, the protons and neutrons in the nucleus can only exist in certain energy levels; see Fig. 2.2. There are separate levels available for protons and neutrons, and each level can only hold two protons or two neutrons. Normally the

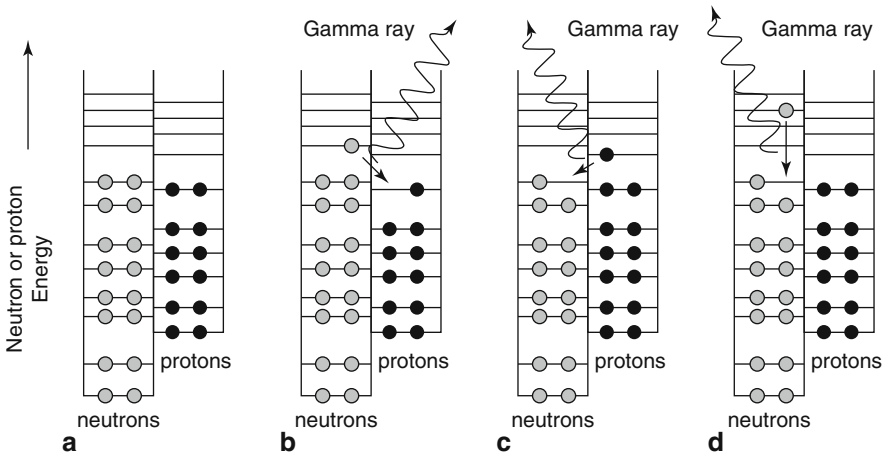

Fig. 2.2 Neutron and proton energy levels in a nucleus. (**a**) Stable – all the lowest levels are filled. (**b**) Beta minus decay – if there is an excess of neutrons, a neutron can convert into a proton emitting energy as a gamma ray. (**c**) Beta plus decay – if there is an excess of protons, a proton can convert into a neutron emitting energy as a gamma ray. (**d**) Isomeric transition – a proton or neutron in an excited state can move down to a lower level emitting energy as a gamma ray

protons and neutrons fill the available energy levels from the lowest upward, and so in a stable nucleus, there are approximately equal numbers of protons and neutrons. Actually, the repulsive force between protons makes them less tightly bound than neutrons, so the proton energy levels are slightly higher than the neutron levels. Therefore, when the levels are both filled to the same energy, the optimum number of neutrons in a stable nucleus is slightly more than the number of protons, as can be seen in Fig. 2.2a. If a nucleus has more than the optimum number of neutrons, then it is energetically favorable for a neutron to convert into a proton so that it can move down to a lower energy level, and in doing so it emits the excess energy as a *gamma ray*. This is the radioactive decay process known as *beta minus decay* (Fig. 2.2b). On the other hand, if a nucleus has more than the optimum number of protons, then a proton can convert into a neutron moving down to a lower energy level and in doing so it also emits a gamma ray. This is known as *beta plus decay* (Fig. 2.2c). It is also possible for a nucleus to have the optimum number of protons and neutrons but for one of them to be at a higher energy level with a vacancy below it. In this case the nucleus is in an excited state, and it decays to the ground state by emission of a gamma ray. This is known as *isomeric transition* (Fig. 2.2d).

2.2 Radioactivity

Radioactivity is a natural process that occurs spontaneously in certain atoms when the nucleus is unstable. It can be explained by the rearrangements of protons and neutrons in the energy levels of the nucleus which have been described above. The

result of a radioactive decay is that energy is released from the nucleus, and this is emitted in the form of radiations which historically were known as alpha, beta, and gamma. *Alpha particles* comprise two protons and two neutrons bound together. They produce a lot of ionization and so are very damaging to human tissue, but they are easily stopped. Alpha emission is not useful in SPECT. *Beta particles* are in fact electrons which produce a moderate amount of ionization in tissue and will only travel a few mm before being stopped. Beta particles are not useful in SPECT, but beta emission sometimes has to be tolerated because it is accompanied by gamma emission which is useful because gamma rays can be detected from outside the body.

2.2.1 Gamma Radiation

Gamma rays are not particles like alpha and beta but are in fact electromagnetic waves, just like radio waves and light waves but with shorter wavelength and higher energy. Although they can behave as waves, it is often convenient to think of gamma rays as small packets of wave energy, or *photons*. In this respect they are no different to X-rays.

The energies of both X-rays and gamma rays are measured in units of thousands of electron volts (keV). One kiloelectron volt is the energy gained by an electron when accelerated through a voltage of 1,000 V. Gamma rays have energies from about 50–1,000 keV, whereas X-rays have energies from about 20–200 keV, so there is great overlap in energy range, and energy cannot be considered an absolute distinguishing feature between them. In fact the only real difference between X-rays and gamma rays is where they originate from; X-rays come from the rearrangement of electrons in atomic energy levels (as illustrated in Fig. 2.1) and gamma rays from the rearrangement of protons and neutrons in nuclear energy levels (as illustrated in Fig. 2.2). So the difference is really just a historical accident of naming conventions. Gamma rays, like X-rays, can easily penetrate human tissue, and therefore, they are useful for SPECT imaging where we want to locate radioactive tracers inside the human body.

2.2.2 Radionuclides

The physical properties of an atom are determined by the numbers of protons and neutrons in its nucleus. A given combination of protons and neutrons is called a *nuclide*. A *chart of the nuclides*, such as Fig. 2.3, shows all possible nuclides with the number of protons, Z, on the vertical axis and the number of neutrons, N, on the horizontal axis, so that each square represents a different nuclide. Some nuclides are

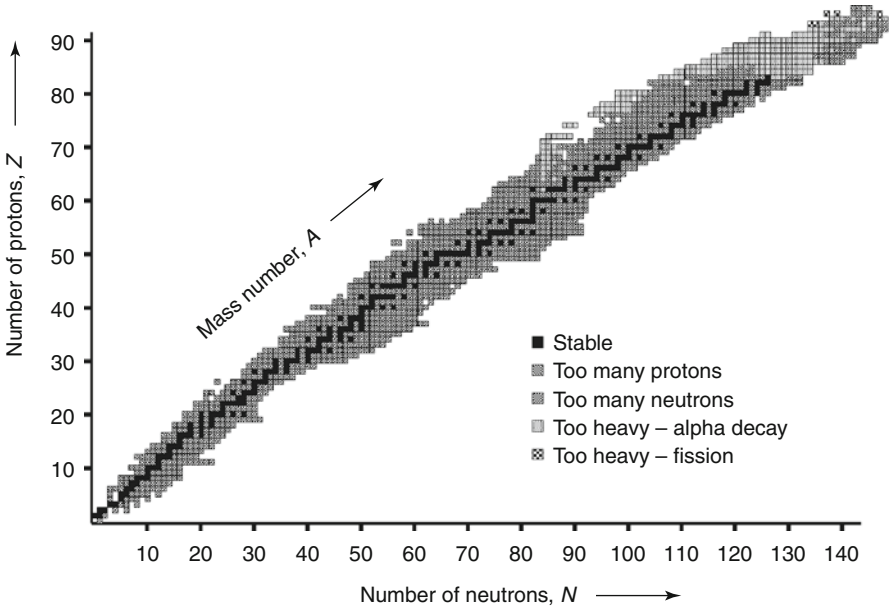

Fig. 2.3 Chart of the nuclides. Each *square* represents a different nuclide with stable nuclides shown in *black* and radionuclides *shaded*

stable, shown by the solid black squares in Fig. 2.3. These occupy a diagonal stripe across the chart of the nuclides, known as the line of stability, where there are slightly more neutrons than protons, as predicted by Fig. 2.2a. Nuclides which lie away from the line of stability, shown by the shaded squares in Fig. 2.3, are unstable because they do not have the optimum combination of protons and neutrons, and these are known as *radionuclides*.

The number of protons in the nucleus determines its electrical charge, which will determine how many electrons can be bound in its electron shells. Since the chemistry of an atom is determined by how it combines with other atoms, which in turn depends only on the number of electrons in its outer shell, all atoms with the same number of protons will be chemically identical. So Z is known as the *atomic number* of the atom, and all atoms with the same Z belong to the same chemical element. Therefore, each row of the chart of the nuclides corresponds to a different element. However, having different numbers of neutrons does not change the chemistry of an atom, and so we call atoms with the same Z but different N *isotopes* of the element. On the chart of the nuclides, all squares on the same row are isotopes of one element. Different isotopes are characterized by their different *mass number* which is the total mass of the atom. Since protons and neutrons have approximately equal mass and the electrons are very light and do not contribute significantly to the mass of the atom (see Table 2.1), the mass number, A, is given by the sum of Z and N, and

this increases diagonally up the chart of the nuclides. Therefore, to distinguish different isotopes of an element, the mass number can be written as a superscript above the chemical symbol representing the element. For example, ^{123}I, ^{124}I, ^{127}I, and ^{131}I are all isotopes of iodine with mass numbers of 123, 124, 127, and 131, respectively. Since iodine always has 53 protons, this means that these isotopes have, respectively, 70, 71, 74, and 78 neutrons in their nucleus. For iodine the optimum number of neutrons is 74, and so ^{127}I is a stable isotope and is not radioactive. However, ^{123}I and ^{124}I are radioactive because they have too few neutrons (or equivalently too many protons), and ^{131}I is radioactive because it has too many neutrons, and these are called *radioisotopes* of iodine.

2.2.3 Electron Capture Decay

Radionuclides with too many protons lie above the line of stability on the chart of the nuclides; see Fig. 2.3. They can decay to a nuclide that is closer to stability by converting a proton into a neutron as already shown in Fig. 2.2c. One way that this can happen is by absorbing one of the orbiting electrons into the nucleus in a process called *electron capture*. The electron combines with a proton to give a neutron plus a *neutrino*. The neutrino has no charge and practically no mass and hardly ever interacts with anything, so it can be ignored for most practical purposes. The result of electron capture is to reduce the number of protons and increase the number of neutrons in the nucleus by one. Therefore, the atomic number reduces by one, but the mass number remains unchanged. For example,

$$^{123}\text{I} + \text{atomic electron} \rightarrow {}^{123}\text{Te} + 159\,\text{keV gamma ray}$$

This shows that an atom of ^{123}I (iodine, with 53 protons and 70 neutrons) can capture one of its own atomic electrons to become ^{123}Te (tellurium, with 52 protons and 71 neutrons). The negative charge of the electron neutralizes the positive charge of the proton to convert it into a neutron. The excess energy is emitted as a gamma ray with an energy of 159 keV. This decay proceeds with a half-life of 13 h. The ^{123}Te produced is stable and does not undergo any further decay.

2.2.4 Beta Plus Decay

An alternative process that can occur with radionuclides that have too many protons is beta plus decay. An example of this is the radioactive decay of ^{124}I

$$^{124}\text{I} \rightarrow {}^{124}\text{Te} + \text{positron} + 603\,\text{keV gamma ray}$$

Here an atom of ^{124}I (iodine, with 53 protons and 71 neutrons) can spontaneously convert into ^{124}Te (tellurium, with 52 protons and 72 neutrons) and a positron, with the excess energy emitted as a gamma ray. The *positron* is sometimes known as a beta plus particle. It is the antiparticle of an electron, which means that it has the same mass as the electron but a positive charge (see Table 2.1). When the emitted positron meets an ordinary electron, the two annihilate each other, resulting in a pair of back-to-back 511 keV gamma rays. This is not useful for SPECT, but it is the basis of positron emission tomography (PET).

2.2.5 Beta Minus Decay

Radionuclides with too many neutrons lie below the line of stability on the chart of the nuclides; see Fig. 2.3. They can decay to a nuclide that is closer to stability by converting a neutron into a proton as already illustrated in Fig. 2.2b. This occurs during the process of beta minus decay (sometimes just called beta decay). An example of this is the radioactive decay of ^{131}I

$$^{131}\text{I} \rightarrow {}^{131}\text{Xe} + \text{electron} + \text{gamma rays}$$

Here an atom of ^{131}I (iodine, with 53 protons and 78 neutrons) can turn into ^{131}Xe (xenon, with 52 protons and 79 neutrons) plus an electron. This decay proceeds spontaneously with a half-life of 8 days. Historically the electron emitted in this process was known as a beta particle. In this case there are two possible beta particles that could be emitted with maximum energies of 333 or 606 keV. These are only quoted as the maximum possible energies because the unseen neutrinos carry off some of the energy so that the beta particles actually have a range of energies from zero up to this maximum. Therefore, the average beta energy is considerably less than these maxima. One of two possible gamma rays is also emitted, with energies of 364 or 637 keV.

2.2.6 Isomeric Transition

Another possible form of radioactive decay is isometric transition, as shown in Fig. 2.2d. This occurs when the nucleus just has an excess of energy, but it does not need to change the numbers of protons or neutrons. It is particularly important in cases of *metastable states*. Normally after a beta decay, the neutrons and protons in the nucleus rearrange their energy levels in a tiny fraction of a second, but sometimes it can be seconds, minutes, or even hours before this occurs. During that time the nucleus is left in an excited state that is said to be metastable because it lasts for an unusually long time. When a metastable state does decay, it only needs to emit a gamma ray, and there are no beta particles emitted, so this is a very clean sort of decay. An important example of a metastable radionuclide is ^{99m}Tc.

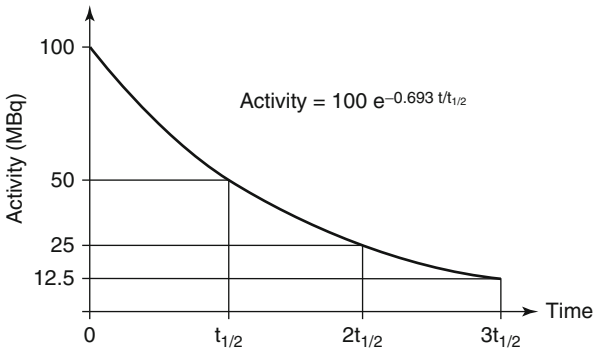

Fig. 2.4 Radioactive decay. The activity decreases exponentially with time with a half-life of $t_{½}$. If the activity is initially 100 MBq, then after one half-life it will have fallen to 50 MBq and after two half-lives to 25 MBq

$$^{99m}\text{Tc} \rightarrow {}^{99}\text{Tc} + 140\ \text{keV gamma ray}$$

Here the "m" in the superscript indicates that ^{99m}Tc is a metastable state of technetium. It decays into ^{99}Tc (without the m) with the emission of just a 140 keV gamma ray. This decay proceeds with a half-life of 6 h.

2.2.7 Activity and Half-Life

The activity of a radioactive source is determined by the number of atoms that disintegrate in a given time. A rate of 1 disintegration per second is given the special SI unit of the Becquerel (Bq). This is a very small amount of activity, so we are more usually dealing with millions of disintegrations per second or megabecquerels (MBq).

The rate of disintegration, and hence the activity, must also be proportional to the number of atoms present, because twice as much radioactive material means twice as many atoms and hence twice as many available to decay each second. However, the decay of a radioactive atom is an entirely random process; it is never possible to say when any individual atom will decay, only how long it will take on average. Therefore, we describe the lifetime of a radionuclide by its *half-life*, which is the average time taken for half of its atoms to decay. Therefore, if we start with a large number of radioactive atoms, after one half-life only half of them will have survived. After another half-life (two half-lives in total) only half of the surviving half will be left (one-quarter of the original). After three half-lives only one-eighth of the original atoms will be left and so on. Therefore, the number of atoms decreases at an ever diminishing rate, following what is mathematically an exponential curve. Since the activity is proportional to the number of atoms present, we can also say that the activity decreases exponentially with time. This is illustrated in Fig. 2.4 which shows that if the initial activity is say 100 MBq, then after one half-life ($t_{½}$) the activity will have decayed to 50 MBq, after two half-lives ($2t_{½}$) to 25 MBq, and after three half-lives to 12.5 MBq.

2.3 Production of Radionuclides

By their nature radionuclides are unstable, and so they will eventually decay away and disappear. Therefore, with the exception of the decay products of some very long-lived heavy elements like uranium, radionuclides do not occur naturally, and so they have to be produced artificially. There are three main methods of doing this, using particle accelerators, nuclear reactors, or radionuclide generators.

2.3.1 Cyclotrons

Radionuclides that decay by electron capture decay or beta plus decay have an excess of protons in their nucleus, and so they can be made by bombarding stable elements with an intense beam of energetic protons. If a proton enters the nucleus of a previously stable target atom, it can turn it into a radioactive atom. Protons are easily produced from ionized hydrogen gas, because hydrogen just has a single proton in its nucleus, so they just need to be given enough energy to overcome the repulsive electrostatic forces and penetrate the nucleus of the target element. This can be achieved using a particle accelerator known as a *cyclotron*. Because protons have an electric charge, they can be accelerated by an electric field, and when they are moving, their path can be bent by a magnetic field. In a cyclotron a beam of protons is made to circulate in a spiral path within a vacuum chamber between the poles of a large magnet. The beam is surrounded by two semicircular hollow chambers known as "dees." An alternating electric field is maintained between the two dees so that the protons are accelerated by the voltage difference as they pass from one dee to the other, and then, half an orbit later, when the electric field has reversed, they are accelerated further as they move back into the first dee. If the voltage difference between the two dees is a few thousand volts, after about a 1,000 orbits, the protons can be accelerated to energies of several million electron volts. As their energy increases the radius of the protons' orbit also increases so that they spiral out towards the edges of the magnet where they hit a suitably placed target.

An example of a radionuclide that can be produced in this way is ^{123}Xe

$$^{127}\mathrm{I} + \text{proton} \rightarrow {}^{123}\mathrm{Xe} + 5\,\text{neutrons}$$

The target is normal stable iodine, and the product is radioactive xenon, which is easy to separate from the target because it is a gas. ^{123}Xe decays by electron capture decay with a half-life of 2 h into ^{123}I, so this is actually a good way to produce ^{123}I which is a useful radionuclide in nuclear medicine.

Because the product is always a different chemical element to the target, cyclotron-produced radionuclides have the advantage that they can be produced *carrier-free*, which means that there are no non-active atoms of the element present in the product. In the above example this means that there are no stable Xe atoms mixed in with the

radioactive ^{123}Xe. Because they are carrier-free, cyclotron-produced radionuclides also have high *specific activity*, which means that there is a large activity per gram of the element in the finished product. However, there is sometimes a possibility of contamination from other radioisotopes of the product that could be produced at the same time. This can be minimized by using pure targets and setting an optimum energy of the proton beam in order to maximize yield of the desired product. However, because maximum proton beam currents are limited, yields can be low, and because cyclotrons are expensive to operate, cyclotron-produced radionuclides tend to be expensive.

2.3.2 Nuclear Reactors

Radionuclides that decay by beta minus decay have an excess of neutrons, and so they can be produced by bombarding a stable target with neutrons. Since neutrons have no charge, they cannot be accelerated like protons, so it is no good using a particle accelerator. However, the lack of charge means that they can more easily penetrate the atomic nucleus, so they do not need to be of high energy. Low-energy neutrons are produced in great abundance in a nuclear reactor, and so neutron irradiation can be achieved simply by placing a target in or near to a reactor. This can result in *neutron activation* of the target. An example of this is the production of ^{24}Na.

$$^{23}\text{Na} + \text{neutron} \rightarrow {}^{24}\text{Na}$$

Here the target is normal stable sodium, and the product is radioactive sodium. Since the product is chemically the same as the target, neutron activation results in a product that has a lot of stable carrier atoms present, so it has a low specific activity.

Another way to produce neutron-rich radionuclides is to make use of *nuclear fission*. Fission occurs in nuclides at the top right of the chart of the nuclides (the checkered squares in Fig. 2.3) which are just too heavy to be stable, so they spontaneously break up into two lighter nuclides. For example, when the nucleus of an atom of ^{235}U absorbs a neutron, it becomes ^{236}U which is so heavy that it immediately breaks up by fission. There are many possible fission fragments that can result, but two of the possible outcomes that are relevant to nuclear medicine are ^{131}I and ^{99}Mo.

$$^{235}\text{U} + \text{neutron} \rightarrow {}^{236}\text{U} \xrightarrow{\text{fission}} {}^{131}\text{I} + \ldots.$$
$$\xrightarrow{\text{fission}} {}^{99}\text{Mo} + \ldots.$$

Unfortunately natural uranium contains less than 1% of the isotope ^{235}U, so the target material first has to be enriched to sufficiently high levels of ^{235}U. Then the enriched uranium target is irradiated inside a nuclear reactor for a few days, and

when it is removed it contains a highly radioactive mixture of many different fission fragments. Because these are all chemically distinct from the original uranium, they can be extracted chemically, although the process has to be done remotely inside a heavily shielded "hot cell" because of the high activity. The resulting products are carrier-free and so have high specific activity, although there may be contamination from other radioisotopes of the element that are produced at the same time and which cannot be separated chemically. Because yields are high and one reactor can irradiate many different targets at once, fission-produced radionuclides are relatively cheap and easy to obtain. However, suitable irradiation facilities are only available in a few research reactors around the world, and these reactors are now very old and prone to breakdown. So it will be necessary for governments to plan replacement facilities in order to maintain the supply of fission-produced radionuclides in the future.

2.3.3 Radionuclide Generators

Cyclotrons and nuclear reactors are expensive facilities which will usually exist at some distance from the nuclear medicine department, and so transport of short-lived products can be a problem. Radionuclide generators provide an ideal way of having a supply of short-lived radionuclides available locally. They rely on a radionuclide with a long half-life (called the *parent*) that decays into another radionuclide with a shorter half-life (called the *daughter*). The most important example of this in nuclear medicine is the ^{99}Mo/^{99m}Tc generator.

$$^{99}\mathrm{Mo} \rightarrow {}^{99m}\mathrm{Tc} + \beta + \gamma^{-}$$

Here the parent ^{99}Mo is obtained by fission of ^{235}U that has been irradiated with neutrons in a nuclear reactor (see above). The ^{99}Mo decays by beta minus emission giving off high-energy gamma rays, so at the molybdenum processing plant, it has to be handled inside a hot cell. Here the ^{99}Mo is adsorbed onto the surface of aluminum oxide particles and loaded into small glass columns which are placed inside a thick lead shield to form the generator. Then the generator is sterilized so that it becomes suitable for patient use. The ^{99}Mo has a half-life of 67 h, so the generator can easily be delivered to distant nuclear medicine departments without undergoing too much decay. The daughter radionuclide, ^{99m}Tc, is metastable with a half-life of 6 h, and after about 24 h an equilibrium is reached where the decay of ^{99m}Tc atoms is matched by production of new ones from decay of ^{99}Mo. In this state sodium chloride solution can be flushed through the column, where the ^{99m}Tc forms sodium pertechnetate solution (Na ^{99m}Tc0$_4$) leaving the ^{99}Mo fixed to the aluminum oxide. This process is called *elution* of the generator, and it delivers useful ^{99m}Tc with very little left in the generator. However, by 24 h later the ^{99m}Tc activity has built up to an equilibrium value again, and the elution can be repeated. Therefore, the generator

can be eluted each day for at least a week, yielding 20 % less each day because of the 67-h half-life of ^{99}Mo. Thus, if a nuclear medicine department purchases a ^{99}Mo/^{99m}Tc generator, each week it can easily obtain a daily supply of $Na^{99m}Tc0_4$ solution. This can be used to prepare many different technetium-labeled radiopharmaceuticals which can be use for SPECT imaging.

2.4 Interaction of Gamma Rays with Matter

When a gamma ray (or an X-ray) passes through a block of material of any type, there is a possibility that it will pass right through without being affected. However, there is also a possibility that it may interact with the material. Interaction occurs mainly through two processes, photoelectric absorption and Compton scattering. These are illustrated in Fig. 2.5.

In the process of *photoelectric absorption*, the gamma ray interacts with an atom of the material and gives up all its energy to ejecting an electron from the atom. This leaves an ionized atom and an energetic electron which is called a *secondary electron*. Photoelectric absorption dominates at low gamma ray energies and in materials with high atomic number. Therefore, materials such as lead provide good shielding against gamma rays, but thicker lead is needed as the gamma ray energy increases. Because it creates an electron vacancy in an inner electron shell, the process will inevitably be followed by emission of characteristic X-rays of the material (e.g., lead X-rays), and so the shielding effect may not be as great as expected.

Compton scattering occurs when a gamma ray (or an X-ray) interacts with one of the outer electrons in an atom which is not so tightly bound. In this process the gamma ray only loses some of its energy, and it continues as a scattered gamma ray with a lower energy and traveling in a different direction. A small loss of energy results in a small angle of scatter and a larger loss of energy results in a larger scatter angle. Maximum energy loss occurs when the gamma ray is completely backscattered and emerges at 180° to its original direction. The loss of energy goes to eject a secondary electron from the atom. Compton scatter dominates at high gamma ray energies and in low atomic number materials, such as human tissue.

If a beam of gamma rays is passed through a block of material, as shown in Fig. 2.5, the process of either photoelectric absorption or Compton scattering results in a steady reduction in the number of gamma rays remaining in the beam; we say that the beam is attenuated. This is analogous to the number of atoms left after radioactive decay, and like radioactive decay, it leads to an exponential fall in the intensity of gamma rays remaining. Therefore, in an analogy with half-life, we measure the *half-value layer* as the thickness of material that will stop half of the gamma rays on average.

Another way to measure the attenuation is through the *attenuation coefficient*, μ (mu), which is the probability that an interaction occurs in each cm of material. For example, the attenuation coefficient for 140 keV gamma rays in human tissue is

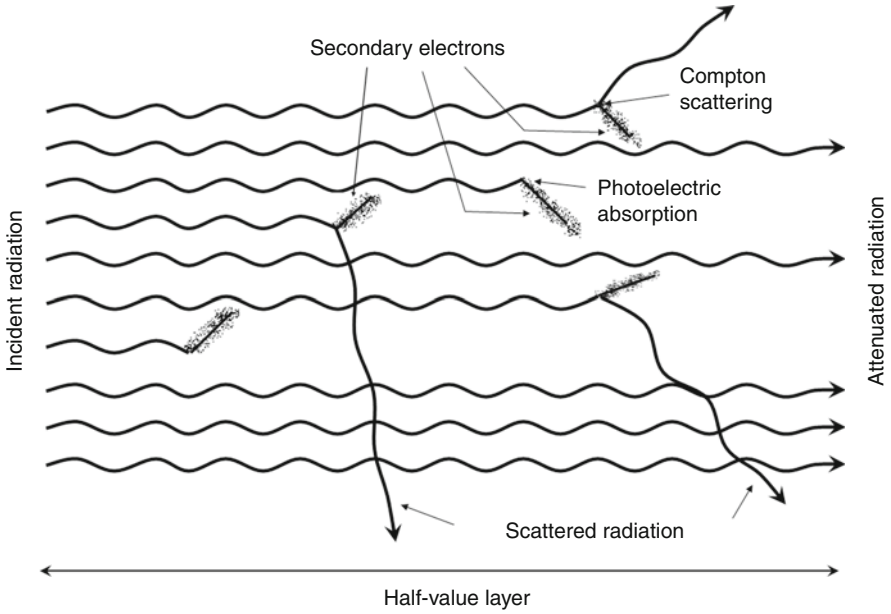

Fig. 2.5 Passage of gamma rays or X-rays through matter. Photoelectric absorption and Compton scattering interactions result in photons being lost from the direct beam, so the intensity of the incident radiation becomes attenuated. The thickness of material needed to halve the number of photons is the half-value layer. The lost energy is passed to secondary electrons which cause ionization of the material

0.15 cm^{-1} which means that a fraction 0.15 (or 15 %) of gamma rays will be lost in every cm of tissue. This corresponds to a half-value layer of 4.6 cm.

Note that gamma rays which pass right through the material without interaction are the ones which can be usefully detected and that these deposit no energy in the material and hence deliver no radiation dose. It is only gamma rays that interact in the material which create secondary electrons, and it is these secondary electrons which go on to cause ionization of surrounding atoms and thus deliver a radiation dose. Thus, optimization of conditions to minimize attenuation not only gives the best transmission of gamma rays for useful imaging but also minimizes the radiation dose to the patient. That is how nuclear medicine procedures can result in low radiation doses despite having the source within the patient.

2.5 Desirable Properties of Radionuclides for SPECT

SPECT studies, like all nuclear medicine procedures, are performed by finding suitable pharmaceuticals that will concentrate in the organ of interest or follow the required metabolic pathway. By labeling these pharmaceuticals with a suitable

radionuclide, we can produce a *radiopharmaceutical* whose presence can be detected by an external detector, which is usually a gamma camera. Therefore, an ideal radionuclide for SPECT imaging should have the following properties:

1. The radionuclide must emit gamma rays and preferably no beta particles, since these will be absorbed in the patient and contribute to the radiation dose without being usefully detected.
2. The gamma ray energy should be high enough to minimize attenuation in the patient. This means that ideally the energy should be greater than about 100 keV.
3. The gamma ray energy should be low enough that it can be efficiently stopped within the detector of a gamma camera. This puts a practical upper limit on the energy of about 300 keV.
4. The gamma ray should preferably have an energy that is low enough to make use of low-energy collimators. These are generally good up to about 200 keV. Above that energy the lead in the collimators has to be made thicker, and the collimator becomes less efficient.
5. The half-life of the radionuclide should be long enough to prepare the radiopharmaceutical and administer it to the patient. It should also be long enough for it not to decay too much during the period of the SPECT scan (possibly up to 30 min) unless there is a system for continuous delivery to the patient.
6. The half-life of the radionuclide should not be so long that it lingers in the patient after the procedure is completed, because this will increase radiation dose.
7. The radionuclide should be readily available at reasonable cost.
8. The radionuclide should be capable of attaching to a variety of useful pharmaceuticals.

Condition 1 means that electron capture decay or isomeric transition is the preferred decay mode. Conditions 2, 3, and 4 mean that the ideal gamma ray energy range is 100–200 keV. Conditions 5 and 6 mean that a half-life of a few hours is ideal. Therefore, ^{99m}Tc (see Table 2.1) is perfect, particularly since it is readily available at any time from a generator that can be kept in stock in a radiopharmacy. Although ^{99m}Tc has slightly difficult chemistry, manufacturers have produced a variety of kits that can be used to label radiopharmaceuticals for imaging a wide range of organs. Therefore, ^{99m}Tc reigns supreme as the most widely used radionuclide for SPECT imaging.

^{123}I (see Table 2.2) is also well suited to SPECT imaging, although it does also have some higher energy emissions which means that it is preferable to use a medium energy collimator which can compromise image quality. ^{123}I also has excellent chemistry because iodine can be incorporated into a wide range of interesting molecules. The main drawback of ^{123}I is that it has to be produced in a cyclotron which means that it has limited availability and is expensive.

Table 2.2 Properties of some radionuclides used in SPECT

Radionuclide	Decay mode	Half-life	Gamma ray energy (keV)	Production method
^{99m}Tc	Isomeric transition	6 h	140	Generator
^{123}I	Electron capture	13 h	159	Cyclotron
^{201}Tl	Electron capture	3 days	70, 167	Cyclotron
^{111}In	Electron capture	2.8 days	170, 245	Cyclotron
^{67}Ga	Electron capture	3.3 days	93, 185, 300	Cyclotron

Chapter 3
Basic Discussion on Electromagnetic Radiation: X-Rays in Relation to SPECT/CT Systems

Peter Hiles

X-rays were discovered by W. C. Roentgen in 1895. They are produced when electrons collide with atoms under appropriate conditions. In this section we will consider the basics of X-ray production and the issues behind the choice of radiation energy for diagnostic imaging.

3.1 X-Ray Spectrum

When high-speed electrons are directed at a target material, the majority of interactions will involve small energy transfers from the high-speed electrons to electrons in the target atoms, which eventually appears as heat. However, occasionally the incident electron can interact with the target atoms to produce one of two types of X-rays, depending on whether the interaction is with the target's orbital electrons or the nucleus.

If the incident electron makes a direct hit on a target electron with sufficient energy, it can cause the electron to be ejected. When the vacancy left by the electron is filled by an electron from a higher energy level (see Chap. 2, Sect. 2.1), energy is released in the form of electromagnetic radiation of a specific wavelength which is distinctive of the target material. This is a so-called *characteristic X-ray*.

If instead the negatively charged electron approaches the positively charged atomic nucleus, the electrostatic force between them will cause the electron to be deflected. If the interaction causes the electron to slow down or brake, the energy loss can result in an X-ray being emitted. These are known as bremsstrahlung or braking X-rays. How much energy is emitted depends on how much energy the electron has retained after previous interactions as well as on the level of deflection that occurs with this particular nucleus. A wide range of photon energies may thus

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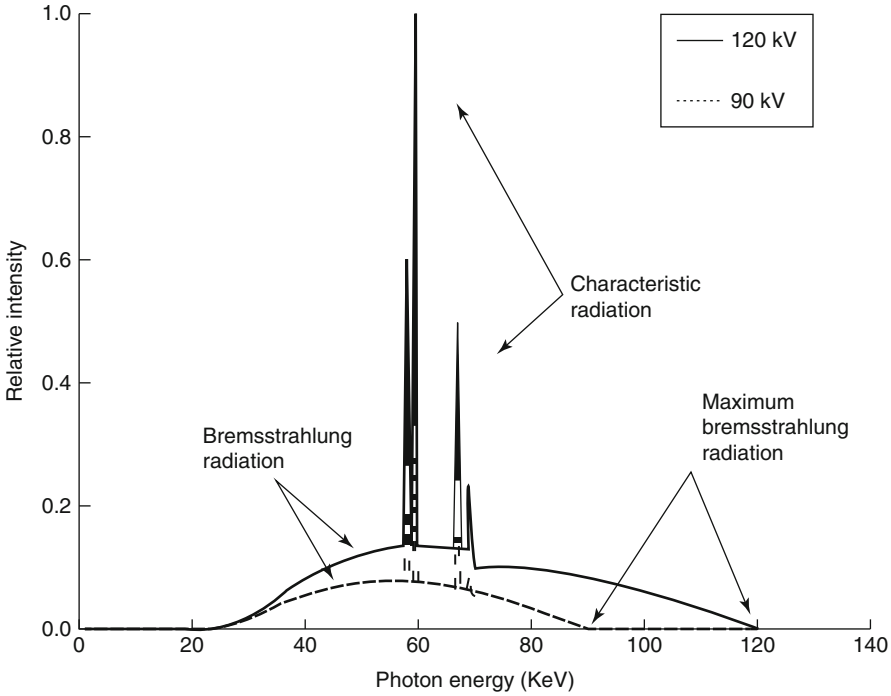

Fig. 3.1 Example tungsten target X-ray tube energy spectrum for applied potentials of 90 and 120 kV, relative to the peak intensity, where the mean photon energies are 53 and 62 keV, respectively, and the area under the *curve* is proportional to the amount of radiation produced. The distributions consist of spikes of characteristic radiation superimposed on a continuous curve of bremsstrahlung radiation (*Curve* adapted from Cranley et al. [1])

be produced by a given target material (Fig. 3.1). However, if the electron undergoes only one interaction with the nucleus and is stopped completely, the whole of its energy appears as bremsstrahlung radiation, which marks the maximum energy limit of the X-ray spectrum. For example, an X-ray tube operating at 120 kV means the electrons striking the target will have gained an energy of 120 keV (see Chap. 2, Sect. 2.2.1) which would produce X-ray photons with a maximum of 120 keV.

The resultant X-ray spectrum (Fig. 3.1) therefore consists of a continuous spectrum (bremsstrahlung) with spikes (characteristic) superimposed on it. This is very different to the spectrum from radionuclides, which consist of specific peaks of radiation energy.

3.2 The X-Ray Tube

The essential elements of an X-ray tube (Fig. 3.2) are a source of electrons (known as the cathode), a target of suitable material (the anode) with which the electrons can be made to collide after having been accelerated to a suitable velocity and an

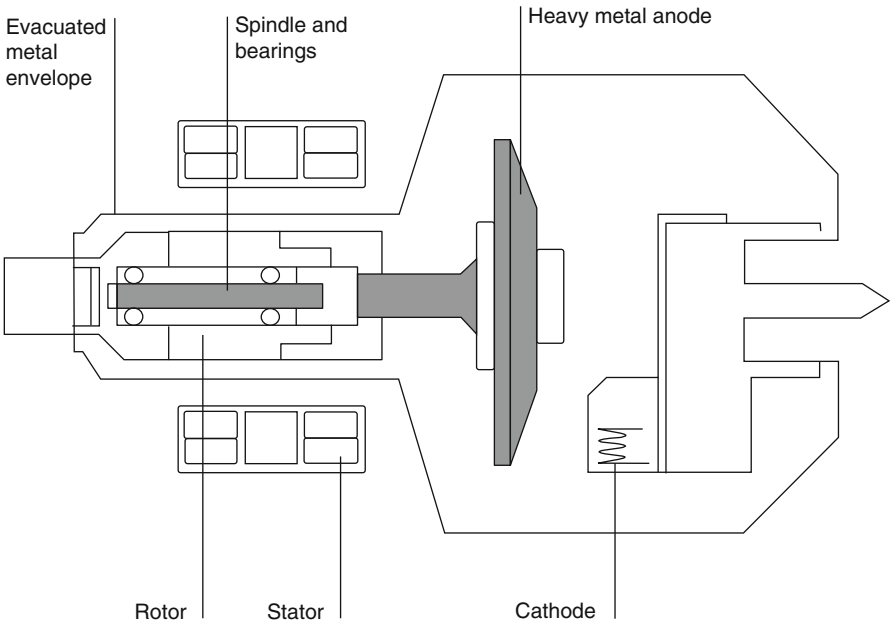

Fig. 3.2 Schematic diagram of a rotating anode X-ray tube

enclosure within which near-perfect vacuum conditions can be maintained to allow the necessary processes to take place.

The construction of X-ray tubes is largely dictated by the fact that X-ray production is a very inefficient process for voltages used in diagnostic radiology, where generally less than 1 % of the energy carried by the electrons is converted to X-rays and over 99 % appears as waste heat and must be removed from the assembly.

X-ray tubes are subject to significant wear and tear and must, in due course, be replaced by new ones.

3.2.1 Cathode

The source of electrons is a helical filament wound from tungsten wire and heated to incandescence by passing an electrical current through it, in a similar manner to a light bulb or projector lamp. The electrons are produced by "evaporation" from a glowing filament, and the amount of electron emission is controlled by varying the heating current. An electrical potential between the cathode and anode accelerates these electrons in the direction of the anode. For diagnostic X-ray production, the potential across the tube may be up to 150,000 V (150 kV).

Associated with the filament and forming part of the cathode assembly is a "focussing cup." This is a metal block that carries the helical filament in a specially

profiled recess which focuses the electrons into a narrow beam as they are accelerated towards the anode target. This ensures that they will be concentrated on a small rectangular area of the anode. This area is the effective source of the X-rays.

3.2.2 Anode

Where the cathode is the source of electrons, the anode is the source of X-rays. The accelerated electrons impinge on the target area of the anode, which is made of a material with both a high melting point and a high atomic number. A high atomic number is necessary because the probability for bremsstrahlung interactions, although low, increases with the atomic number of the target area of the anode. A high melting point ensures that the target does not vaporize with the high energy of the beam.

The basic target material is usually tungsten (typically alloyed to rhenium to improve its mechanical strength), because of its high atomic number (74) and ability to withstand very high temperatures (melting point 3,370 °C). The tungsten disc is backed with a material such as molybdenum (melting point 2,620 °C and specific heat capacity twice that of tungsten) to increase the thermal capacity of the anode for a given weight. Alternatively graphite could be used which, weight for weight, has five times the heat storage capacity of molybdenum.

The disc is mounted on a spindle carried on a bearing assembly and the rotor of the anode drive motor (induction motor). If the bearing lubricant is in the form of a film of liquid metal, then this also offers the possibility of direct cooling of the anode rotor. The heavy metal disc can be made to rotate at speeds of up to 9,000 rpm (150 Hz). During rotation, including acceleration from start-up to final steady speed, significant forces can arise which could distort the rotor. Therefore, considerable care is taken to provide suitable suspension of the anode assembly.

The typical diagnostic X-ray tube has its anode in the form of a rotating compound metal disc (Fig. 3.3). The electrons bombard a small area on the face of the disc (the target), near its periphery, but because the disc is rotating, the heating effect is distributed around an annulus or "focal track." This enables much heavier loads to be applied to a bombarded area of a given size. Anode disc diameters can vary from 70 to 200 mm, where the larger anode diameter at the same rotational speed offers a longer track length, and so the heat generated is spread over a greater area. Larger discs therefore can take higher loading.

During an exposure, the heat from the focal track is distributed by conduction over the anode disc. If this is not followed by immediate and effective dissipation of the heat from the anode disc, the next exposure can only take place if it is still possible for the heat to be distributed over the anode. As soon as the whole of the anode disc is hot, additional exposures must be postponed until the heat has been dissipated from the anode by radiation or conduction. Thus, depending on the heat capacity of the anode, a limit number of exposures can be made in rapid succession,

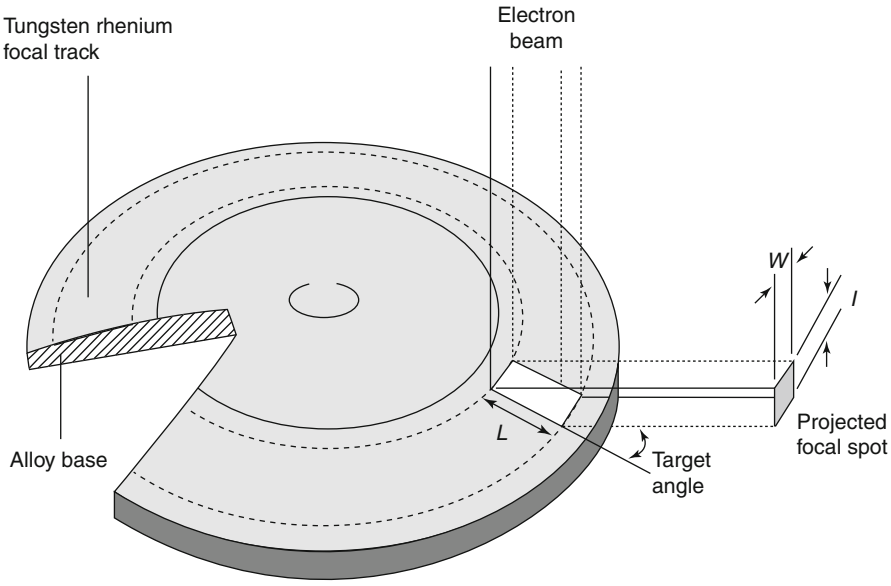

Fig. 3.3 Anode disc with a cutaway to highlight the base material. The target area is shown as a *white rectangle* and the focal track by *dashed lines*. The diagram also illustrates the relationship between the actual focal length on the anode, *L* (and width, *W*), and the projected or effective focal length, *l*

after which the tube needs a cooling-off period. If the radiation is to be produced continuously, the load is determined by the rate at which the heat is dissipated from the anode.

The disc loses its heat by radiation to its surroundings and by conduction through the spindle and bearings. In a rotating anode tube, the loading depends on the speed of rotation, the length of the focal track, and on its width. Thermal loading is also dependent on the mass of the anode and the level of suitable cooling of the anode. This last item has a significant impact on the long-term loadability which is an essential requirement for X-ray CT applications.

To consider the tube loading issues for CT, consider the following example. In CT the X-ray tube is mechanically rotated about a patient (360° rotation) while it is emitting radiation. This rotation typically takes 0.5 s and is repeated several times in quick succession. Therefore, to enable continuous use of the CT scanner without enforced waiting times while the tube cools down requires efficient dissipation of heat from the tube. In addition, the tube also has to meet special mechanical requirements with respect to the stability of the focal spot during the motion of the gantry. Consequently modern CT tubes contain a large anode disc with a diameter of 200 mm which provides extremely efficient heat radiation and a very high heat storage capacity. When this is coupled with direct cooling of the rotating anode, waiting for the tube to cool is virtually unknown even with heavy tube loads.

3.2.3 Focal Spot

As illustrated in Fig. 3.3, the X-rays do not originate from a single point on the anode surface, but from a rectangular (target) area. The maximum X-ray intensity that can be obtained from such a small area on the target is limited, as we have seen, both by the material and the rate of cooling of the target. One way to increase the intensity of the beam is to project the X-ray beam at an angle to the plane of the target (Fig. 3.3). In this way it is possible to use a relatively large area on the target for X-ray production which is then projected into a smaller area towards the image receptor. It is the size of the projected area in a defined direction that is known as the *effective focal spot*, the length of which is dependent on the target angle. This is an important parameter since the size of the effective focal spot can affect an imaging systems ability to resolve small details in the acquired image.

3.2.4 Vacuum Tube

The two main components of an X-ray tube (anode and cathode) are enclosed in a vacuum-tight vessel (Fig. 3.2) in order to allow the electrons free movement towards the anode, without interacting with air particles. The tube envelope also provides the necessary electrical insulation.

In the past, diagnostic X-ray tubes had a glass envelope or "bulb." However, more recently, designs using metal envelopes with bonded ceramic insulating parts have become common, which have greater strength and heat conduction properties.

The envelope is evacuated during manufacture, and elaborate processes are applied to ensure that during the life of the tube there will be no breakdown in the vacuum.

3.2.5 X-Ray Tube Housing

The whole X-ray tube is fitted into a metal container or *tube housing*. This contains the stator coils, which are part of the induction motor used to rotate the anode assembly within the evacuated tube envelope. The housing also includes connectors for the high-voltage and stator cables, bringing the electrical supplies to the assembly. An X-ray port (or window) provides a defined exit for the useful beam of X-rays. The housing is lined with lead to reduce the amount of radiation escaping, other than through the window, to below specified limits (known as *leakage radiation*). The housing is also required to absorb and dissipate the heat radiated from the anode during X-ray production. This tube assembly can be connected to an external heat exchanger, which ensures an adequate flow of oil and radiates the acquired heat energy to the surrounding air.

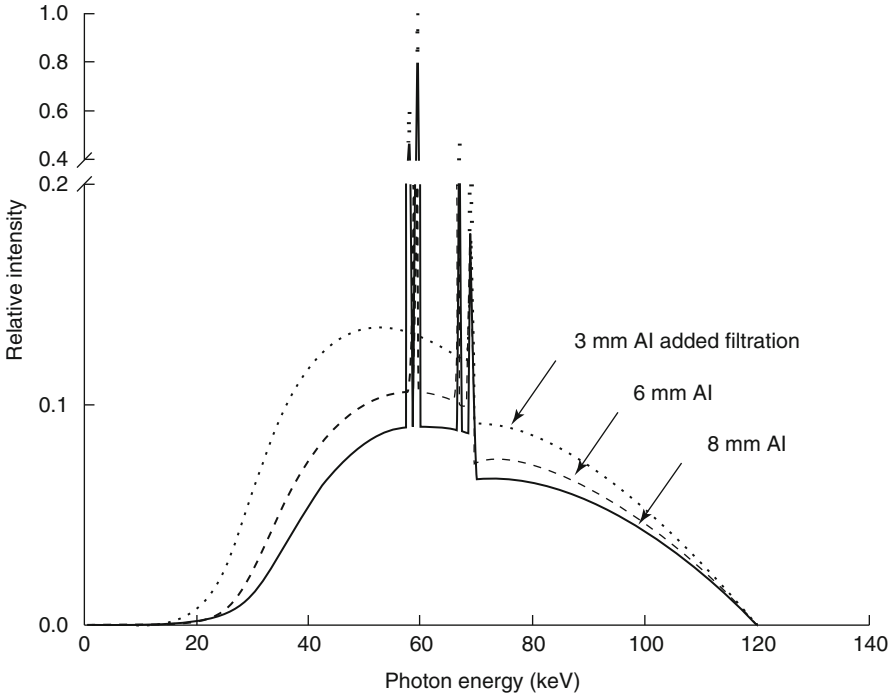

Fig. 3.4 Example tungsten anode (target) X-ray tube energy spectrums for applied potential of 120 kV and added filtration of 3, 6, and 8 mm of aluminum (*Al*) relative to the peak intensity, where the mean photon energy increases from 56 to 62 keV while the peak energy is unchanged at 120 keV. The vertical axis has been broken in order to highlight the differences in bremsstrahlung intensity with filtration (*Curves* adapted from Cranley et al. [1])

3.2.6 Filtration

An X-ray image is formed by the radiation reaching the image receptor as a result of passing into or through the body of the patient. The generation of X-rays, however, produces large components of low-energy ("soft") radiation which, if allowed to reach the patient, would be absorbed near the skin surface. This would add to the radiation dose received by the patient without contributing to the formation of the image. In order to remove these unwanted parts of the X-ray spectrum, small sheets of metal or filters can be placed between the tube and the patient to intercept the X-ray beam. Figure 3.4 illustrates the effect of adding increasing amounts of *filtration*. The filters absorb the "soft" components preferentially, which alters the proportion of high- to low-energy photons in the spectrum, and so the mean photon energy or *effective energy* of the beam increases. Adding filtration thus has the effect of "hardening" the beam.

The filtration is usually expressed in terms of the equivalent thickness of aluminum (Al) that, when compared under test conditions, would exhibit the same absorption effect as the filters under consideration. To meet safety requirements,

a certain minimum amount of filtering material must always be present, and this must be at least partly in the form of irremovable material in the X-ray tube assembly (known as *inherent filtration*). Additional filters may be fitted to produce particular beam characteristics, tailoring the spectrum to the imaging requirements. Therefore, in CT, where whole body imaging is envisaged, the X-rays produced must have sufficient energy to penetrate even the most attenuating areas of the patient, and so filtration equivalent to 6–10 mm of aluminum can be used.

3.3 High-Voltage Generator

Within the X-ray equipment as a whole, the high-voltage generator is the part that supplies and controls all the electrical inputs to the X-ray tube assembly. It is required to provide power to the X-ray tube (in the range 10–200 kW), output voltage in the range 20–150 kV, and current ranging from 10 to 1,000 mA.

Modern generators include various sensors to determine the instantaneous values of X-ray tube voltage and current. If these differ from the desired or selected values, they can then be adjusted in real time. This tight control of exposure accuracy and reproducibility means modern X-ray generators are extremely reliable.

Due to the variations in power supplies (single phase, three phase, 50/60 Hz), which can be present at an X-ray installation, generator designers have had to provide a solution which is independent of the commercial line input. This has led to the almost universal introduction of *frequency converter generators*. This includes a power supply that converts the single- or three-phase input into an intermediate DC source and inverter that converts the DC input into an AC pulse, which is then input into a high-voltage (HV) generator to supply the X-ray tube. This design has led to the construction of compact and cost-effective generators, which has proved particularly important for CT scanners, where the generator is incorporated into the rotating part of the gantry.

3.3.1 Applied Potential (kV)

In Fig. 3.1, the applied tube potential has been increased from 90 to 120 kV. This increases both the energy (curve extended towards the higher photon energies) and amount (area under the curve) of X-rays emitted from the tube. For diagnostic radiography, the amount (or intensity) of X-rays produced approximately increases with the square of the kilovoltage.

The choice of kilovoltage determines the object (patient) penetration and image quality (see Sect. 3.4). Low kilovoltages can improve the distinction between soft tissues and so improve the contrast in the image. Higher kilovoltages increase the effective energy of the X-ray beam, thereby increasing the penetration (see also Sect. 3.2.6). In addition, since the beam intensity increases approximately with kV^2,

the overall amount of radiation reaching the image detector also increases. For example, increasing the kilovoltage from 100 to 120 kV (20 % increase) can increase the X-ray beam intensity by 50 %, as well as increasing the effective energy.

Therefore, in CT, where suitable and sufficient penetration is required of the whole body, a tube potential of 120–140 kV is generally used for adults. For pediatrics, where less penetration is required, this can be reduced to 80–100 kV.

3.3.2 Applied Current (mA)

The number of X-rays produced is dependent on the number of electrons which strike the target, and the number of electrons depends directly on the tube current (mA) applied. The greater the mA, the larger the number of electrons that are produced, and consequently more X-rays are produced. However, it should be noted that varying the tube current (mA) does not influence the energy of the X-rays emitted.

3.4 Tailoring the X-Ray Spectrum for Diagnostic Imaging

X-ray imaging consists of detecting the radiation transmitted through the body, where the ability to distinguish between tissues in the image is due to differences in radiation attenuation. Choosing a suitable X-ray spectrum to acquire diagnostic images can therefore be explained by considering how X-rays are attenuated as they pass through a material. The reduction in the intensity, or attenuation, of the radiation beam as it traverses matter, was described in Sect. 2.4.

For a beam of monoenergetic X-rays of intensity I_0, incident on a uniform material (in density and atomic number) of thickness x, the transmitted X-ray beam intensity I_T is given by exponential equation:

$$I_T = I_0 e^{-\mu x}$$

where μ is the *linear attenuation coefficient*, which indicates how much of the photon beam is attenuated as it passes through a given thickness of a material.

Figure 3.5 illustrates the variation in μ with energy for three tissue types. This figure shows that the lower the photon energy, the greater the attenuation difference between tissues. For example, at 50 keV, the difference between the attenuation of bone and muscle is 0.34 cm^{-1}, which is reduced to approximately a third (0.12 cm^{-1}) at 100 keV or 15 % at 500 keV.

The energy range required for diagnostic CT X-ray imaging is a compromise between obtaining suitable contrast (attenuation) differences between tissues and having a beam with sufficient penetration. Although, as we have seen, the diagnostic X-ray spectrum is not monoenergetic but consists of a range of photon energies,

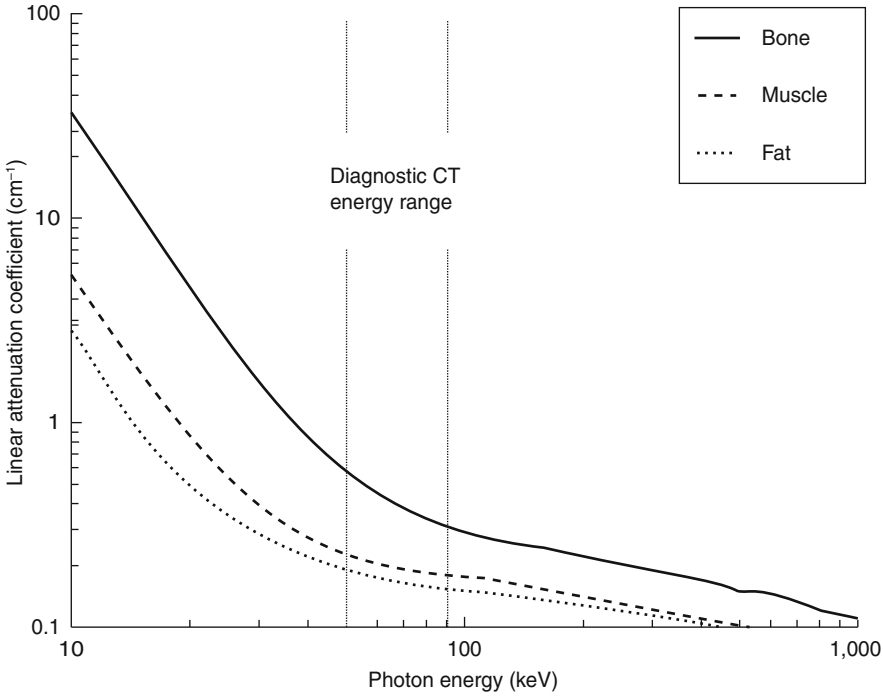

Fig. 3.5 Linear attenuation coefficients for three tissue types plotted on a logarithmic scale. The *vertical dotted lines* indicate the mean photon energy range for diagnostic CT X-rays (50–90 keV). The graph is based on data in Johns and Cunningham [2]

the tube potential (kV) and added filtration are designed to give a mean photon energy for CT imaging of between 50 and 90 keV. Taking the example of the 120 kV spectrum in Fig. 3.1, which has an average photon energy of 62 keV, the μ values for bone, muscle, fat, and water at this energy are 0.44, 0.21, 0.18, and 0.20 cm^{-1}, respectively. While the difference between bone and fat is significant, the difference between muscle and water is clearly quite small. To enhance small differences between different tissue types, the intensity scale (called the *CT number*) used in the reconstructed CT image is defined by

$$\text{CT number} = \frac{\mu - \mu_{\text{water}}}{\mu_{\text{water}}} \times K$$

where μ_{water} is the linear attenuation coefficient of water and K is a constant.

The CT number is often given in Hounsfield units (HU), where the constant K is assigned the value of 1,000 and, by definition, water has a CT number of 0 HU. The CT numbers for the three materials considered of bone, muscle, and fat would then be +1,170 HU, +35 HU, and −120 HU, respectively. This provides a significant contrast range which is exploited in X-ray CT imaging.

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Chapter 4
Gamma Camera SPECT

Richard S. Lawson

4.1 Principles

The name Single Photon Emission Computed Tomography (SPECT) describes exactly what the technique entails. "Single Photon" refers to the fact that it utilizes radionuclides that decay with the emission of a single gamma ray photon. This distinguishes it from Positron Emission Tomography (PET) which utilizes radionuclides that decay by positron emission resulting in production of a pair of annihilation gamma ray photons. "Emission" refers to the fact that radiation is emitted from the patient, as opposed to transmission where radiation would be transmitted through the patient. "Computed Tomography" refers to that fact that computers are used to reconstruct tomographic cross sections through the patient.

Nuclear medicine images are usually acquired with a gamma camera, which is a device that can detect gamma rays emerging from the patient over an area of typically about 50 × 40 cm. Thus, a gamma camera image shows the distribution of radioactivity in the patient, and this reflects the distribution of radiopharmaceutical, which in turn maps the function of the organ or system being imaged. However, the gamma camera, like an ordinary photographic camera, does not record any depth information, and so it only produces a two-dimensional (2D) image of what is actually a three-dimensional (3D) distribution within the patient. The process of *tomography* overcomes this limitation by acquiring gamma camera images from several different directions around the patient; these are called *projections*. Using a computer, these 2D projections can then be reconstructed to reproduce the full 3D distribution of activity within the patient. The results are usually displayed as cross sections taken across the patient (transaxial slices) or along the patient viewed from in front (coronal slices) or viewed from the side (sagittal slices).

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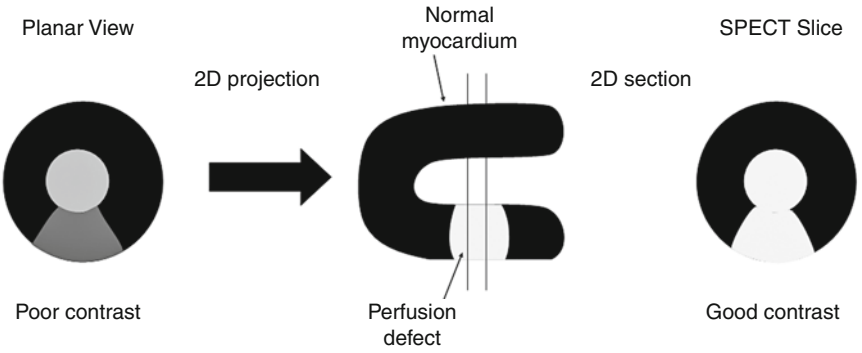

Fig. 4.1 Illustrating how SPECT gives better image contrast than planar imaging for a cold lesions. The planar view of a perfusion defect within normal myocardium shows poor contrast due to overlying normal myocardium. The SPECT slice through the defect shows good contrast

SPECT imaging has several advantages over standard planar nuclear medicine imaging. Most obviously the availability of three-dimensional information allows the position of any abnormality to be more accurately located. It also allows activity to be more accurately quantified, because a 3D region of interest can be drawn around an object of interest which isolates it from surrounding background. On an ordinary planar image, a 2D region of interest will always include background from activity that lies between the object and the gamma camera and also from behind the object. Moreover, many gamma rays fail to be detected by the gamma camera because they are absorbed in the intervening tissues. This attenuation means that the detected counts can be a significant underestimate of true activity in a region of interest. It is difficult to do much about this attenuation on a planar image because the depth of the source is not known. However, with SPECT, because full 3D data is available, the depth of every source is known, and so a correction can be made for lost counts. This is known as attenuation correction (see Sect. 4.4.2).

A third advantage of SPECT over planar imaging is less obvious but probably more important. If the object under study contains a region of low activity within surrounding areas of normal activity, this is called a "cold lesion." If such a cold lesion is viewed with a gamma camera, then the area of the abnormality will have tissues with normal activity lying in front and behind it. Therefore, on the planar image, these normal tissues will be seen as superimposed on the abnormal tissue. The difference between normal and abnormal areas in an image is measured by *image contrast*, and so this superposition will reduce the contrast of the lesion. However, if the object is viewed with a SPECT image, then any slice through the cold lesion will show good contrast because the overlying normal tissues will appear in different slices and they will no longer be superimposed. Thus, SPECT is particularly useful for imaging organs where cold lesions may be present within normal tissue. Myocardial perfusion imaging and brain perfusion imaging are good examples of this. In these two applications, planar imaging is no longer performed, and SPECT is standard practice. Figure 4.1 illustrates how SPECT can improve the

visualization of a perfusion defect within the wall of a myocardial perfusion scan. The planar view of the heart seen end on is a 2D projection, which sums the full thickness of the myocardium, and so the defect is hard to see because the contrast between normal and abnormal areas is poor. However, the SPECT slice creates a 2D section through the heart, and this shows the defect easily because it has good contrast.

4.2 Basic Instrumentation

This section will give an overview of the basics of equipment used to acquire SPECT images. It is not intended to be a comprehensive description, and the reader is referred to other works for further details [1, 2].

4.2.1 The Gamma Camera

The gamma camera was invented by Hal Anger in 1958 [3], and so it is sometimes known as the Anger camera. Although the technology has been improved over the years, the basic design is still used in most conventional gamma cameras. Figure 4.2 illustrates the principles of operation of a modern gamma camera.

The back and sides of the gamma camera are covered with lead shielding to prevent unwanted gamma rays from entering in this direction. The front face of the gamma camera is covered with a *collimator*, which is essentially a lead plate about 4 cm thick with many thousands of parallel holes through it, each about 2 mm in diameter. Gamma rays emerge from the patient in all directions, but only those that pass through the collimator holes can reach the detector behind it. Therefore, only gamma rays that are traveling perpendicular to the collimator face will be detected, and gamma rays traveling at an angle will be stopped. Without the collimator the gamma camera would detect gamma rays coming from all directions, but as it is unable to determine their direction, it would not know where they came from. Thus, the collimator is essential for forming an image, and so it serves the same purpose as the lens of a photographic camera.

Immediately behind the collimator is a single large *scintillation crystal* which can be up to about 50 × 40 cm in area and 1 cm thick (Fig. 4.2a). The crystal is usually made of sodium iodide with a small amount of thallium doping added, referred to as NaI(Tl). This material has the property that when a gamma ray hits the crystal, there is a high probability that it will be absorbed by photoelectric absorption (see Chap. 2) and the absorbed energy causes the crystal to emit a flash of light. This process is known as *scintillation*, and so the gamma camera is also sometimes called a scintillation camera. NaI(Tl) is hygroscopic, so the crystal has to be hermetically sealed to prevent it from absorbing moisture from the air. This is done with a thin metal encapsulation over the front surface which faces the collimator and a

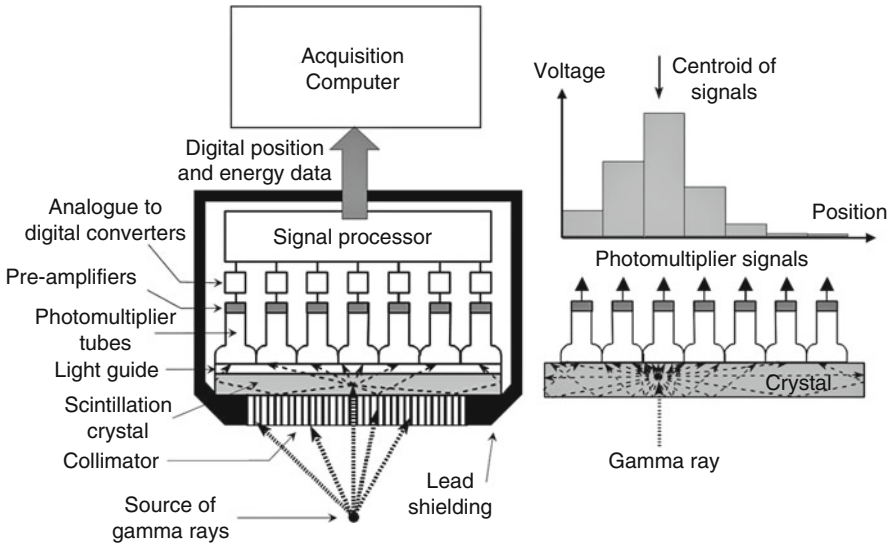

Fig. 4.2 Principles of operation of the gamma camera. (**a**) Basic components of a modern camera. (**b**) Determination of the event location from the centroid of all photomultiplier tube signals

transparent glass window, known as the *light guide*, over the rear face. The inner surface of the metal encapsulation is covered with reflecting material so that all the scintillation light is reflected out through the light guide. A typical gamma ray that interacts with the crystal will produce several thousand light photons, and although these are in the visible light range, they are too faint to be easily detected by eye.

The scintillation light is collected and amplified by an array of *photomultiplier tubes* which are optically coupled to the light guide (Fig. 4.2a). There may be up to 100 photomultiplier tubes in a large gamma camera, and each one is usually about 5 cm diameter. The photomultipliers are evacuated glass tubes containing a series of electrodes known as dynodes. The front face of each tube, called the photocathode, is coated with a material which emits electrons when light hits it. The electrons are accelerated toward the first dynode which is held at a high positive voltage, so they hit it with sufficient energy to release several more electrons. These electrons are accelerated toward the second dynode where even more electrons are released, and so on, until after 10 dynodes, the number of electrons can be amplified by a factor of about one million. This makes photomultipliers sufficiently sensitive to be able to detect even single light photons, and so the few thousand light photons in a scintillation event are easily detected. A pulse of several million electrons reaching the final anode represents a small electric charge, but when this is passed through a preamplifier, it then becomes a signal that is large enough to be processed by the electronics.

In a modern gamma camera, the signal from each preamplifier is digitized by an *analogue to digital converter* and passed to a digital *signal processor* (Fig. 4.2a). This compares the sizes of the signals from each photomultiplier tube in order to

determine the position of the scintillation event. Tubes that are closer to the event will detect more light and so produce a bigger signal, and so the centroid of the signals (analogous to the center of gravity of an object) will be a first estimate of the scintillation position (Fig. 4.2b). The position estimate is improved by applying a *linearity correction* which allows for variations in light collection efficiency across the crystal. In this way, the position can be determined to within about 3 or 4 mm, which is known as the *intrinsic spatial resolution* of the camera. The sum of all the photomultiplier signals is a measure of the total light emitted, and this is proportional to the energy deposited by the gamma ray. The signal processor also applies an *energy correction* to allow for variations in light collection efficiency, and the final energy can be determined with a resolution of about 10 %. If the gamma ray was stopped by photoelectric absorption, then the deposited energy will be equal to the energy of the gamma ray itself and so the energy signal can be used to distinguish between different radionuclides.

The data about the position and energy of each detected gamma ray is passed to the acquisition computer which builds up an image by storing events in an image matrix which may be 64 × 64, 128 × 128, or 256 × 256 pixels in size. Only events that have an energy within some predetermined energy window are accepted. The image may be terminated after a preset time or preset number of counts. When the acquisition is finished, the number of counts in each pixel may be adjusted to compensate for variations in camera sensitivity across the field of view. This is known as *sensitivity correction* or sometimes *uniformity correction*.

4.2.2 Collimators

As already mentioned, the collimator is an essential part of the gamma camera; without it, there can be no image. The most commonly used collimators have parallel holes. Their purpose is to only let through gamma rays that are traveling perpendicular to the detector face. In practice, because of the finite size of the holes, a small range of angles will be admitted, and this results in a blurring of the image. The blurring is measured by the *collimator resolution*, which is the minimum distance between two point sources at which they can just be separated in the image. Figure 4.3 shows that the collimator resolution depends on the size of the holes and that it gets worse the further away the sources are from the collimator. Unfortunately the collimator is very inefficient, and typically only about 0.01 % of gamma rays get through the collimator. This means that a source of 1 MBq (one million disintegrations per second) would only give a count rate of 100 cps. Thus, the *collimator sensitivity* would be quoted as 100 cps/MBq. It can be made more sensitive by using larger holes which will let through a wider range of angles, but this will make its resolution even worse. Therefore, collimator design is a compromise between resolution and sensitivity. High-resolution collimators are designed to have small holes to give good resolution, but this means that they have poor sensitivity. High-sensitivity collimators have larger holes, but this gives then worse resolution.

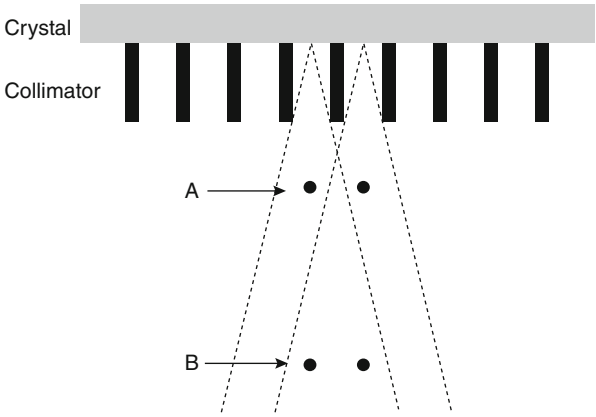

Fig. 4.3 Illustrating why collimator resolution gets worse with increasing distance. The two sources at *A* which are close to the collimator can be resolved because they are each "seen" through different collimator holes. The two sources at *B* are further from the collimator, and they are not resolved because they can both be "seen" through the same two collimator holes

General-purpose collimators have medium-sized holes and therefore moderate sensitivity and resolution.

The overall *system resolution* of a gamma camera image depends on both the intrinsic resolution of the crystal and the resolution of the collimator that is fitted according to the formula

$$(\text{System Resolution})^2 = (\text{Intrinsic Resolution})^2 + (\text{Collimator Resolution})^2$$

Figure 4.4 shows how the system resolution varies with distance for three different collimators. For all collimators, best images (smallest resolution) will be obtained with the patient as close as possible to the collimator. At 0-cm distance, all collimators give similar system resolution because this is dominated by the intrinsic resolution of the crystal, but as distance increases, the high-resolution collimator maintains better resolution than the others. Figure 4.4 also shows that sensitivity does not change with distance but that the high-resolution collimator has the worst sensitivity (smallest cps/MBq).

4.2.3 SPECT-Capable Gamma Cameras

In order to perform SPECT studies, the gamma camera must be able to acquire projection images from a wide range of angles around the patient. This is usually achieved by having the patient lie on a narrow bed and mounting the gamma camera detector (or head) on a gantry that can rotate all the way around the bed to any angle. Since it is important to keep the detector as close to the patient as possible (for best resolution – see above), the head must also be able to move in and out to different radial distances. Because the detector head contains a lot of lead, it is very heavy and so the gantry must be sturdily designed to make sure that the heads can move

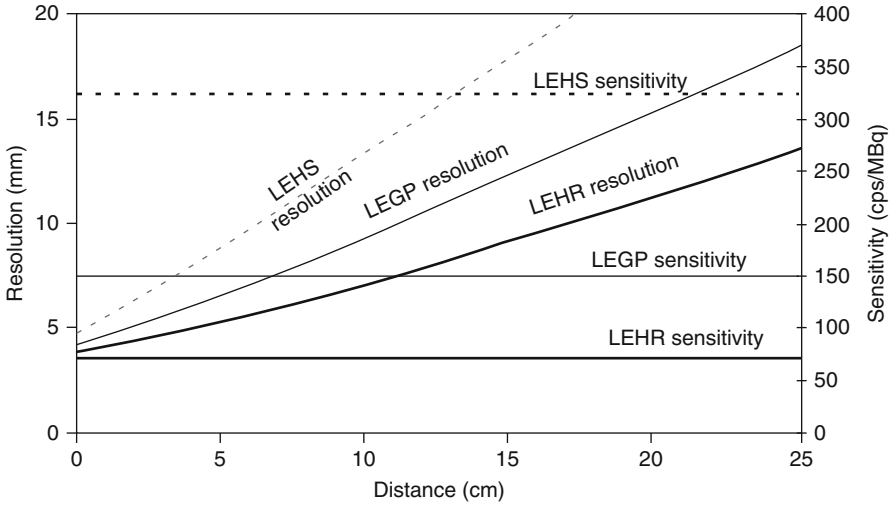

Fig. 4.4 System resolution and sensitivity of a gamma camera using three different collimators. *LEHR* low energy high resolution, *LEGP* low energy general purpose, *LEHS* low energy high sensitivity

smoothly and accurately to any desired angle and radial position. Two typical SPECT-capable gamma cameras are shown in Fig. 4.5.

The gamma cameras shown in Fig. 4.5 each have two separate detectors. This is a common feature of gamma cameras that are to be used for SPECT because it allows two different projection images to be acquired simultaneously. This advantage can be put to use in one of three ways. Firstly, the total time required for a study could be halved, which offers better patient acceptability and also greater patient throughput. This could be useful in studies such as myocardial perfusion SPECT where patients are required to lie with their arms above their head which can be uncomfortable. Secondly, the total acquisition time could be maintained, but twice as many counts could be obtained using two detectors. This could be useful in studies such as ^{111}In white cell SPECT where there is little activity in the patient, and so images can be very noisy unless enough counts are acquired. Alternatively, the acquisition time could be maintained and a better resolution collimator used. Because high resolution implies less sensitivity, this method would give the same number of counts, but the images would have improved resolution. This could be useful in studies such as bone SPECT where there is adequate activity in the patient, but improved resolution enables smaller details to be seen in the image. Thus, whichever approach is used, two heads are almost always better than one in the context of SPECT. In fact, some specialized systems even incorporate three heads.

If a gamma camera has two or more heads, then they may be positioned in several different ways. Figure 4.5a shows the two detectors positioned 180° apart, which is sometimes called "H-mode." H-mode is convenient when projections can be acquired all the way around the patient, because a 180° rotation of the gantry will

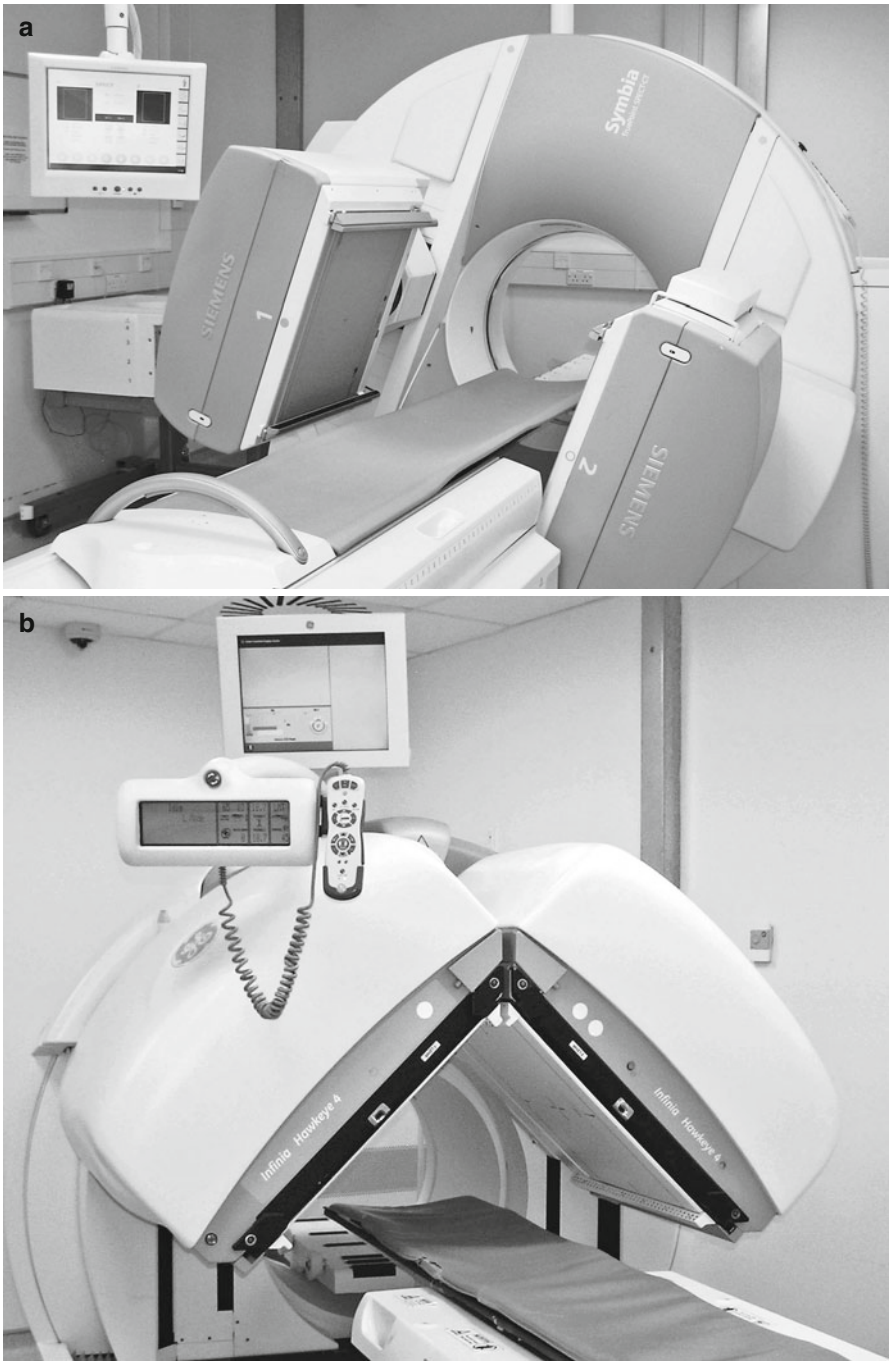

Fig. 4.5 Typical SPECT-capable gamma cameras. (**a**) Siemens Symbia, shown with the two detectors in H-mode. (**b**) GE Infinia, shown with the two detectors in L-mode

cover a full 360° arc when the two detectors are taken together. H-mode is used for most SPECT studies of the head or body. Figure 4.5b shows the two detectors positioned 90° apart, which is sometimes called "L-mode." L-mode is used when projections only need to be acquired halfway around the patient, because a 90° rotation of the gantry will enable the two detectors to cover a 180° arc. This mode is used for SPECT studies of the heart, because the heart lies close to the left anterior chest wall, and so projections from the right and posterior views give very poor quality images. They have poor resolution (because of the increased distance to the heart) and few counts (because of the increased attenuation by body tissues). General-purpose SPECT systems are usually capable of positioning the detectors in both H-mode and L-mode so that any type of SPECT study can be performed.

4.2.4 Dedicated Cardiac Systems

If a SPECT system is only going to be used for cardiac studies, then the detectors can be considerably smaller than they would be for a general-purpose SPECT system. The detectors can be fixed permanently in L-mode, and they only need to move around the left side of the patient, which greatly simplifies the gantry design. In addition, the two detectors can be housed within a single lead shield. This reduces the dead space that results where the two detectors touch, enabling them to get closer to the patient. Therefore, manufacturers have produced a range of dedicated cardiac SPECT cameras. These offer significant advantages for cardiac SPECT, but since they cannot do anything else, they are only suitable for departments that perform a large number of myocardial SPECT studies.

Although most SPECT-capable gamma cameras image the patient in a horizontal position, there are a few dedicated cardiac systems that image the patient in a reclining or an upright position. This can improve patient acceptability as well as reducing the overall floor area required. However, because the heart and other organs lie differently when the patient is upright, normal appearances can be different with these cameras [4].

4.2.5 Hybrid SPECT/CT Systems

Nuclear medicine images obtained with a SPECT system are good for showing organ function, but they give very limited anatomical information, partly because of poor resolution and partly because normal features are often not visualized as they have little uptake of radiopharmaceutical. On the other hand, X-ray images obtained with a CT scanner give no functional information but are excellent for visualizing anatomy. Therefore, a hybrid SPECT/CT system which combines these two complementary modalities in one machine allows both the abnormal function and its anatomical location to be obtained in a single imaging session. This is much better

than taking the SPECT and CT images on separate machines, because with the hybrid system the patient does not have to be taken off the imaging bed between the SPECT and CT images. This should ensure correct registration between the two images which is essential if they are to be used for localization. A SPECT/CT system avoids the difficulties of having to resize, reorient, and realign separate SPECT and CT images that have been taken on different occasions, and it is the main reason why these hybrid systems have become so popular. The first commercial SPECT/CT system was the Hawkeye from GE Medical Systems [5] which became available in 2000. The Hawkeye was a purpose-designed low-dose CT scanner built into the gamma camera gantry, but since then, all the major manufacturers have produced their own versions of hybrid systems.

One advantage of a SPECT/CT system is that the CT image can be used for attenuation correction of the SPECT image (see Sect. 4.4.2). Attenuation correction only needs a very low-resolution CT image which can be acquired with a low radiation dose. But anatomical localization requires a better quality CT image which therefore involves a higher radiation dose. In some cases, a high-quality CT image may also be needed for full diagnosis, and this necessitates an even greater dose. Therefore, manufacturers offer a variety of different systems with choices of CT quality and corresponding radiation dose. Some systems simply add a standard multi-slice CT scanner at the back of a gamma camera gantry, while others integrate specially designed CT systems into the gamma camera. The various features of these systems will be discussed in detail in Chap. 6.

4.2.6 Solid-State Systems

The conventional gamma camera design that has been described so far utilizes a single large scintillation crystal and an array of photomultiplier tubes for each detector. However, in the early 2000s, manufacturers began to introduce new designs of gamma camera that did not use photomultiplier tubes. These are often referred to as solid-state system systems because the vacuum tube photomultipliers have been replaced by semiconductor devices. One advantage of this change is that solid-state systems should be more reliable in operation; photomultipliers require a stable high voltage supply and are susceptible to magnetic fields and deteriorate with age, while semiconductor devices are more robust and do not require a high voltage supply. However, a more radical change results from the fact that dispensing with rather large photomultiplier tubes has resulted in more compact detectors, and this has enabled a variety of novel designs to be developed.

The Anger camera design utilizes a single large NaI(Tl) scintillation crystal and then determines the event position within the crystal by analyzing the distribution of light collected by an array of photomultiplier tubes. The image is then arbitrarily divided up into individual pixels for convenience in storage and display. In contrast, all solid-state detector designs utilize detectors that are inherently pixelated because they consist of an array of many individual small detector elements, each with their

own electronic circuits. The intrinsic resolution of such a pixelated detector is determined by the size of each element, which could be less than 1 mm if required. However, in practice there is no advantage in using such small elements because the collimator will still contribute a resolution of several millimeters. Therefore, commercial detectors tend to have elements about 3 mm in size and so give a very similar resolution to the conventional Anger camera.

In principle, it is possible to produce pixelated detectors of any size by using sufficient detector elements, but since the cost increases in proportion to the number of elements, commercial implementations have started with small field of view systems. Therefore, solid-state detectors were initially introduced for small planar imaging systems used for applications such as mammography. Their first application to SPECT came with dedicated cardiac systems where the small field of view was not a problem and the compact design enabled the detectors to keep close to the heart.

Some solid-state cameras use thallium-activated cesium iodide, CsI(Tl), as the scintillator material. This material only produces half as much light from each scintillation when compared with NaI(Tl), and the scintillation light is green rather than blue. The light is picked up by small photosensitive diodes which are glued to the back of each detector element. In fact, the detection efficiency of the photodiode is better than a photomultiplier, and this makes up for the reduced light output, so that CsI(Tl) has an energy resolution that is slightly better than NaI(Tl). This means that narrower energy windows can be used, which can reduce scatter in the image (see Sect. 4.4.3). Another advantage of CsI(Tl) is that it is not hygroscopic, and so it does not have to be hermetically sealed like NaI(Tl), and also it is not easily cracked. These properties make it easy to fabricate into small pixelated arrays.

Another material that is favored for solid-state detectors is cadmium zinc telluride, CZT. This is an alloy of cadmium telluride and zinc telluride which forms a semiconductor material. It is not a scintillator, so when a gamma ray hits the material, it does not give off light, but instead it releases electrons that can be collected directly by an amplifier circuit. Therefore, there is no need to employ a photomultiplier or even a photodiode. By eliminating the inefficient light emission and collection stage, CZT detectors can produce good signals that are less noisy than those from a scintillator. This gives the possibility of very good energy resolution, but it is expensive to produce high-quality material, and so in practice, energy resolution may only be slightly better than NaI(Tl).

4.3 SPECT Acquisition

Suggested acquisition protocols for a range of clinical SPECT studies are detailed in Chap. 13. This section will therefore just discuss the principles behind the choice of suitable parameters.

In order to acquire SPECT data with a rotating gamma camera, for most studies it is desirable to obtain a sequence of images from a full 360° arc around the patient. Therefore, if the system has two heads, these are positioned in H-mode (see

Sect. 4.2.3) and the gantry only needs to rotate through 180°, but if the system only has one head, the gantry will have to rotate through 360°. Data is usually acquired in step and shoot mode, where the heads remain stationary for a few seconds while each view is acquired and then acquisition pauses while the gantry rotates to the next view position. The user can specify the time per view and the number of views to be acquired, or equivalently the angular step between views. Typically the angle between views is 3°, and so a total of 120 views will be acquired over a full 360° arc. For a double-headed system, there will be 60 views from each head. If the angle between views is set to 6°, then only 60 views will be required, 30 from each head.

As already discussed in Sect. 4.2.3, for SPECT studies of the heart, it is usual to position the two heads in L-mode and only acquire images over a 180° arc, from right anterior oblique, through anterior and left lateral to left posterior oblique, because this gives the best-quality images of the heart. If views are still obtained every 3°, then a total of 60 views (30 from each head) will be required, but each view can be acquired for twice as long as the corresponding 360° arc so the total acquisition time is the same.

One other important parameter that the user must define is the size of the image matrix that will be used to store each projection image. This will usually also be the size of image matrix that is used for the final reconstructed sections, and this determines the pixel size. For planar imaging, when the best possible resolution is required, it is desirable to use the smallest possible pixels, and so a matrix of 256 × 256 pixels is common. For a large gamma camera with a field of view about 50 cm, this means that each pixel will be about 2 mm across. This is much less than the system resolution of the camera at a typical organ depth, and so the small pixels do not further degrade the resolution of the acquired image.

However, the situation is rather different for a SPECT acquisition because the acquired images are not viewed directly but have to undergo computer processing in order to produce the final reconstructed images. In SPECT reconstruction, noise is a big problem, and noise gets worse as the counts per pixel reduce. Small pixels will give fewer counts in each pixel, and so there is no point in making pixels any smaller than they need to be. The general rule for digital imaging is that in order to preserve resolution the pixel size should be less than one-third of the system resolution, but in SPECT, this is often relaxed to suggest a pixel size between one-third and one-half of the system resolution [6, 7]. So, for example, if we use a high-resolution collimator to image a small part of the body, like the brain, where the center of the organ might be about 15 cm from the collimator face, then the system resolution would be about 9 mm (see Fig. 4.4). Therefore, a pixel size of 4 mm would be adequate, and we would use a 128 × 128 matrix to achieve this. However, if we are imaging the heart, although some views can get close and give good resolution, we also have to include many other views where the heart is further away, so the average distance of the heart from the collimator might be 25 cm, and the resolution would therefore be only about 13 mm. Therefore, a pixel size of about 6 mm would be appropriate, and this could be achieved with a 64 × 64 matrix.

The angular separation between views will clearly also affect the resolution of the final image, and so it is useful to consider why a step angle of 3° is often used.

The individual projection images sample the source activity along radial lines that can be thought of as being like the spokes of a bicycle wheel. With this analogy, the angular resolution is determined by the spacing of the spokes at the rim of the wheel where they are furthest apart. In the case of brain SPECT, the diameter of the head is about 20 cm, and so "spokes" positioned every 3° would make them 5 mm apart at the "rim." This is similar to the pixel size, which we chose to be 4 mm, and therefore the spatial sampling resolution and angular sampling resolution are nicely matched. For myocardial SPECT, the diameter of the part of the image containing the heart is probably about 25 cm, so 3° spokes would be about 6 mm apart at the rim, which again matches the chosen pixel size.

For best possible resolution, it is important to keep the collimators as close as possible to the patient. Therefore, even if a simple circular rotation is used, the radius of the orbit must be adjustable to suit individual patients. In fact, since patients tend not to be exactly circular in cross section, manufacturers have gone to considerable trouble to design gamma camera gantries that can rotate the detectors in a variety of noncircular orbits. In all cases, the aim is to keep the collimator close to the patient at all angles while still allowing sufficient gap to allow safe rotation without danger to the patient.

For some parts of the body, such as the brain, a simple circular orbit may be adequate, but for others, such as the chest, an elliptical orbit will be much better. However, for myocardial SPECT where only a 180° orbit is needed, a half-circular orbit can still be a reasonable fit to the body provided that the patient is placed in the optimum position. Some systems achieve noncircular orbits by getting the operator to position the camera as close as possible to the patient at four set positions around the body (e.g. anterior, posterior, and both laterals) and then moving the camera in an ellipse through these points. Others use an automatic contouring system where the camera senses the proximity of the patient's body, for example, by using a series of infrared beams across the face of the collimator. Then the gantry can rotate in a circular orbit, but the detectors can automatically vary the radius as they move round in order to keep at the optimum distance from the patient.

4.4 SPECT Reconstruction

Once projection images of the patient have been acquired, the final task is to deduce what was the actual distribution of radiopharmaceutical in the patient that resulted in these projections. This is the process of SPECT reconstruction, and it is carried out digitally by a computer. For many years, the traditional method of reconstruction was the process of filtered back projection, but modern systems offer the alternative of iterative reconstruction which has many advantages. Full details of both methods can be found in a recent publication from the Institute of Physics and Engineering in Medicine [8], so this section will just give an overview of the main points.

The process of SPECT acquisition is actually a projection process, because each view forms a 2D representation of a 3D activity distribution in the patient. Each

pixel in the projection image represents the sum of activity along a line running from that pixel all the way through the patient. This line is known as the *projection ray*. In reality the projection ray is not a perfect line but a diverging cone because of the poor resolution of the collimator, but a correction for this can be applied using resolution recovery (Sect. 4.4.5). Also, not all activity in the patient contributes equally to the projection image, because gamma rays from deeper sources are less likely to be detected because of attenuation by photoelectric absorption and Compton scattering in the patient. Attenuation correction can be used to correct for this (Sect. 4.4.2). Conversely, some gamma rays can be detected even though they did not originate within the projection ray because they can undergo Compton scattering in the patient. This can be corrected by scatter correction (Sect. 4.4.3). Therefore, in describing SPECT reconstruction, it is simplest to start with the assumption that the projection ray is a perfect line and then apply corrections for each of these effects.

4.4.1 Filtered Back Projection

Simple back projection works on the premise of reversing the forward projection process that created the projection images during acquisition. The process starts with an image matrix to represent each reconstructed transaxial slice. Initially the transaxial image matrix is empty, that is, all pixels are zero. Then for each pixel of the first projection view, the corresponding projection ray is traced across the transaxial image, and the counts from the projection image pixel are spread equally across all the pixels of the transaxial image through which the projection ray runs. Since the projection image contains no depth information, there is no way to know from where along the projection ray the counts actually originated, so spreading them equally is the simplest starting point. This process is repeated for each projection view so that stripes of counts are added into the transaxial image from each projection angle.

If the actual object consists of a single point source, then under ideal circumstances (no loss of resolution, no attenuation, and no scatter) each projection image will contain just one hot pixel. Each projection image will therefore contribute a single hot stripe across the transaxial image. The back-projected stripes from all projection images will all cross at one point and build up a hot spot in the reconstructed transaxial image as illustrated in Fig. 4.6a. It can be seen that this is far from being a correct reconstruction of the source (which should be a single hot pixel) as there is a star-shaped burst of counts all around the true location. This blurring of the reconstructed image is an inevitable consequence of the back-projection process, but fortunately, the shape of the blurring is well defined, and so it can be removed by appropriate mathematical manipulation. It can be shown [8] that the blurring follows a mathematical $1/r$ function, that is, the counts fall off inversely as the distance from the object. To remove this $1/r$ blurring, it is necessary to filter the image with a sharpening filter, and the appropriate filter turns out to be a *ramp filter*,

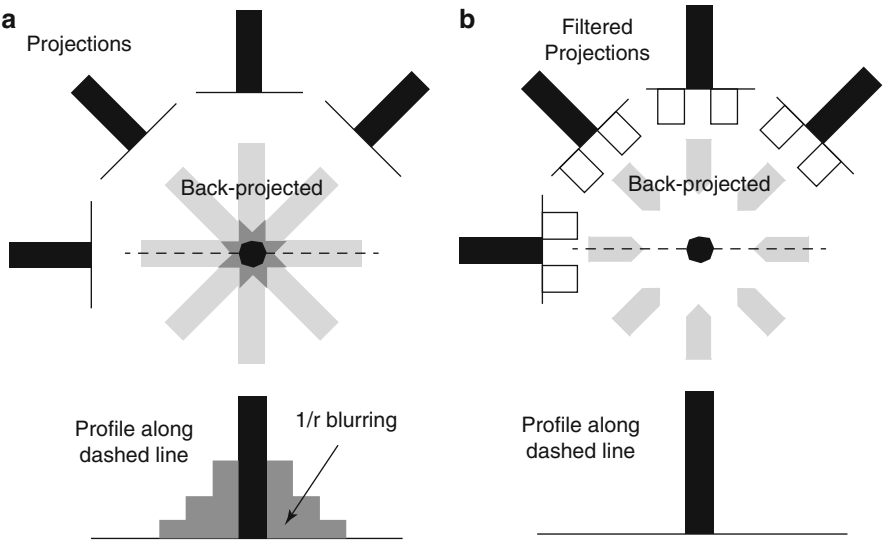

Fig. 4.6 Illustrating the principle of reconstruction of a point object using filtered back projection. For simplicity, only four projections are shown. (**a**) Simple back projection without any filtering gives a starburst pattern with a profile that shows blurring of the central image. (**b**) Filtered back projection adds some negative values to the projections. When these are back projected, they can erase the *star* and remove the 1/*r* blurring

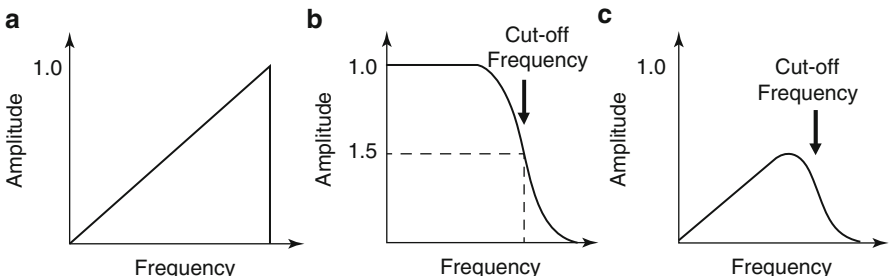

Fig. 4.7 SPECT reconstruction filters. The frequency response shows how the filter affects different spatial frequencies in the image. (**a**) Ramp filter. (**b**) Smoothing window. (**c**) Combination of ramp with smoothing window

that is, one where the amplitude increases linearly with frequency as illustrated in Fig. 4.7a. If the projection images are filtered with this ramp filter before back projection, then the single hot pixel gets some negative counts added to the pixels on either side, and when these are back projected, they can wipe out the starburst effect as shown in Fig. 4.6b.

Unfortunately, although the ramp filter is theoretically perfect for removing the 1/*r* blurring, it has the property that it amplifies high spatial frequencies in the image. Now high spatial frequencies correspond to rapid changes in counts from one pixel

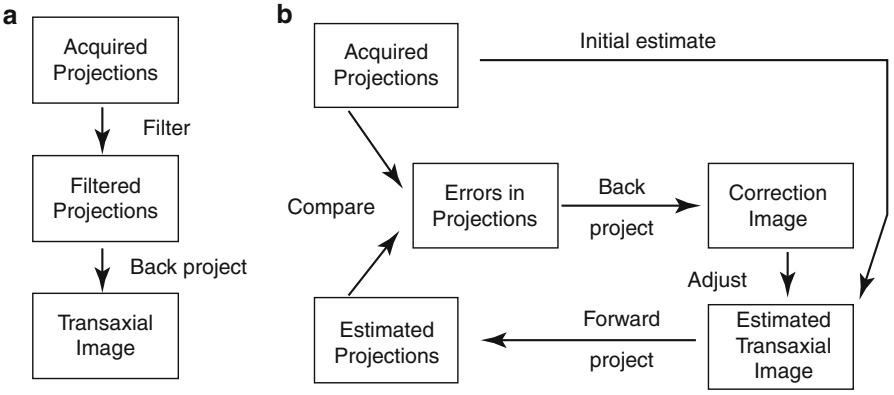

Fig. 4.8 SPECT reconstruction methods. (**a**) Filtered back projection. (**b**) Iterative reconstruction

to another but, since nuclear medicine images have poor spatial resolution, such changes are more likely to be random noise rather than genuine image structure. Therefore, the ramp filter has the undesirable effect of amplifying noise in the images. So, in order to keep the image noise under control, it is necessary to also apply a *smoothing filter*. This takes the form of a low-pass filter that reduces the amplitude of high frequencies while leaving low frequencies unchanged as illustrated in Fig. 4.7b. The combination of ramp filter and smoothing window gives the combined filter illustrated in Fig. 4.7c. At low frequencies, this starts off like the ramp filter in order to remove some $1/r$ blurring, but at high frequencies, it rolls off in order to avoid amplifying image noise.

Although the above process has been described for a single point source, a real extended source can be considered as a collection of many point sources, and so applying the process to every pixel will result in the whole object being reconstructed. The process of SPECT reconstruction using filtered back projection is therefore very simple. It is illustrated in Fig. 4.8a. The acquired projection images are first filtered with a ramp filter modified by a suitable smoothing window. Then the filtered projections are back projected to form a reconstructed transaxial slice. This process can be repeated for a range of transaxial slices to build up a complete stack of sections. The 3D pixels in this stack are called voxels. Other views, such as sagittal, coronal, or oblique, can then easily be produced by reordering these voxels without any further reconstruction. Therefore, filtered back projection is a quick and easy way to reconstruct SPECT data, which explains why it has been popular for many years. Its biggest difficulty is in choosing the correct smoothing filter.

The smoothing filter can take many different mathematical forms, and descriptions such as "Butterworth," "Hanning," and "Hamming" refer to different shapes of the frequency response curve. The most important parameter of any smoothing window is its cutoff frequency. This is the frequency at which its amplitude falls to a value of 0.5 (or sometimes 0.7) as shown in Fig. 4.7b. A low cutoff frequency will get rid of noise in the image but will leave a smooth image with poor resolution, so

small objects will not be visible. A high cutoff frequency will preserve the original image resolution but at the expense of a very noisy image, so low contrast objects will not be visible. So when using filtered back projection, it is vital to select the appropriate cutoff frequency for the smoothing filter in order to obtain a compromise between noise and resolution. The optimum filter will depend on the number of counts in the image, the pixel size, the system resolution, and the size and contrast of the object, and so it may well be different for each clinical application and for each individual user since acquisition conditions will vary from one department to another. In fact, the trade-off between resolution and noise is a subjective decision, and so it can ultimately come down to the personal preference of whoever is going to report the images. Therefore, it is not possible to give definitive rules about what filter should be used in any particular application.

4.4.2 Attenuation Correction

Attenuation is loss of gamma rays by interaction within the patient before they reach the gamma camera. This can occur because of either photoelectric absorption or Compton scatter (see Chap. 2). Figure 4.9a shows some possible gamma ray paths from a patient containing an organ that has accumulated radiopharmaceutical. Gamma ray "A" has reached the detector without interaction and passed through the collimator to be detected. This is a good event. Gamma ray "B" undergoes photoelectric absorption in the patient and so fails to reach the detector and therefore is lost from the image. Gamma ray "C" interacts in the patient by Compton scatter and so continues in a different direction and with lower energy. Because it is now traveling in the wrong direction, it is stopped by the collimator, and so it too is lost from the image. Gamma rays "B" and "C" both contribute to attenuation because they are gamma rays that should have been detected but were not. Clearly this loss of gamma rays will cause the activity in the patient to be underestimated, and the effect will be greater the deeper the source organ is in the patient. Attenuation correction has to deal with this problem, and it should therefore lead to a more accurate quantification of activity.

The process of filtered back projection takes no account of attenuation, and so attenuation correction must be applied as an additional step. This can be done in one of two ways: either as a preprocessing step before filtered back projection or as a post-processing step after filtered back projection.

The most common preprocessing method is due to Sorenson [9]. Two opposing projection views (i.e. ones that are 180° apart) are first averaged and then corrected for an average attenuation depth which depends only on the total thickness of the patient and the average attenuation coefficient of the patient's body. The patient thickness can be estimated from an orthogonal projection, and the attenuation coefficient is assumed to be uniform throughout the patient. Because it relies on averaging two opposing views, the Sorenson method can only be used when a full 360° acquisition has been obtained. The method is quick and easy, but it is rather crude and is rarely used nowadays.

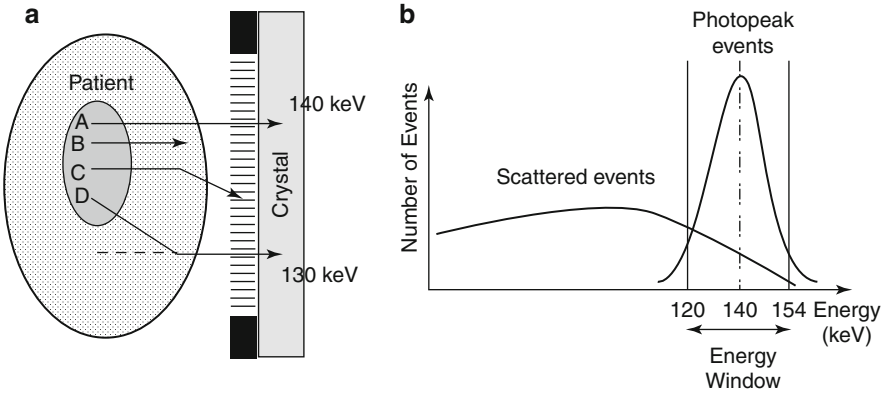

Fig. 4.9 Illustrating attenuation and scatter. (**a**) Some gamma ray paths. (**b**) Corresponding contributions to the detected energy spectrum. "A" is a good event detected in the right place and within the energy window. "B" is a gamma ray lost by photoelectric absorption, and "C" is lost by Compton scatter. *B* and *C* both contribute to attenuation which results in an underestimate of true activity. "D" is a gamma ray that undergoes Compton scatter but still results in an event within the energy window. It is therefore detected in the wrong place which results in a loss of contrast in the image

A commonly used post-processing method is due to Chang [10]. It works on the transaxial slice after this has been reconstructed by filtered back projection. First, the edge of the patient's body is determined either by taking a low threshold on the image counts or by getting the user to position an ellipse around the body outline. Then for every pixel within the body, its distance to the edge of the body along a projection ray is determined, and the attenuation over this distance is calculated. This attenuation is then averaged over all acquired projection angles to give an average attenuation factor for that pixel. The process is repeated to give an attenuation factor for every pixel, and this is used as a scaling factor to increase the reconstructed counts appropriately. This method has the advantage that it will work with either a 180° or 360° acquisition and it is more accurate than the Sorenson method, so it is the one most commonly available.

Both the Sorenson and Chang methods have the limitation that they assume that the attenuation coefficient is uniform throughout the patient's body. This assumption is reasonably good in the head and in the abdomen, but it is inadequate for the chest because of the presence of lungs. At 140 keV (the gamma ray emission energy of ^{99m}Tc), the attenuation coefficient of most body tissues is about 0.15 cm^{-1} and bone is about 0.19 cm^{-1}, so they are not enormously different, whereas lungs have an attenuation coefficient close to 0.0 cm^{-1}. So neither the Sorenson nor Chang method will work in the chest, and therefore, an alternative method that can incorporate nonuniform attenuation is required for studies like myocardial perfusion SPECT. There is an iterative version of the Chang method which can deal with nonuniform attenuation, but this is rarely used, and commercial SPECT systems usually only offer nonuniform attenuation correction as part of iterative reconstruction which will be discussed in Sect. 4.4.4.

To implement any nonuniform attenuation correction method, we need to know the actual attenuation coefficient within the patient. Since every patient will have lungs of different size and shape, it is necessary to obtain a patient-specific map showing the distribution of attenuating material in each individual patient. Since internal organs will move when the patient moves, it is important to obtain this attenuation map with the patient in the same position that they were in for the SPECT study. To this end, manufacturers have developed a variety of systems for acquiring an attenuation map while the patient is still on the imaging bed.

Several manufacturers have produced attenuation correction devices using one or more sources of gadolinium-153 (^{153}Gd). This radionuclide has a long half-life and emits gamma rays of 103 keV whose attenuation properties are similar to those of ^{99m}Tc at 140 keV. If the ^{153}Gd source is placed on one side of the patient and the gamma camera on the other, then the "shadow" cast by the patient creates a transmission image similar to a plain X-ray. If the transmission image is repeated at many different angles, these can be reconstructed into a cross-sectional map of attenuation coefficients, rather like a CT image. So we now need to make a distinction between the *transmission images* that are used to create an attenuation map and the *emission images* which are the standard SPECT study, using ^{99m}Tc, for example.

An advantage of using ^{153}Gd is that it only gives a very low radiation dose to the patient. The ^{153}Gd sources are housed in shielded containers on the gamma camera gantry and only exposed when they are in use. The ^{153}Gd energy of 103 keV is sufficiently different to the ^{99m}Tc energy of 140 keV that both transmission and emission images can be acquired simultaneously by using two different energy windows on the gamma camera. This has the advantage that the total acquisition time is not extended, and it also guarantees that the patient position is the same for both emission and transmission images. Because the number of transmitted gamma rays is low, the transmission images are rather noisy, but they are perfectly good enough to distinguish lungs from tissue and hence to create an adequate attenuation map. However, because ^{153}Gd has a half-life of 8 months, the sources do need to be replaced at intervals. If the sources are not replaced, then as they decay the attenuation maps can be so noisy that they become useless.

Once manufacturers started producing hybrid SPECT/CT systems, where CT scanners are integrated into the gamma camera, it became obvious that the CT scan provided a very convenient way to produce the attenuation map. Therefore, with SPECT/CT, there is no longer any need for external ^{153}Gd sources. The advantage of using a CT scan to create the attenuation map is that the quality of the image is much better than that from ^{153}Gd, and so it is accurate enough to be able to extrapolate to any other energy from 70 keV (^{201}Tl) all the way up to 511 keV (PET). Of course a CT scan will give the patient a much higher dose than the ^{153}Gd, so the additional dose must be justified against the benefit of a more accurate attenuation correction. By using CT parameters that produce low-quality, low-resolution studies, the patient dose can be kept to around 1 mSv and still produce CT images that are adequate for attenuation correction purposes, even though they cannot be used for diagnosis.

A standard CT scan produces images with pixel values that represent tissue density in terms of the CT number measured in Hounsfield units (HU). In order to

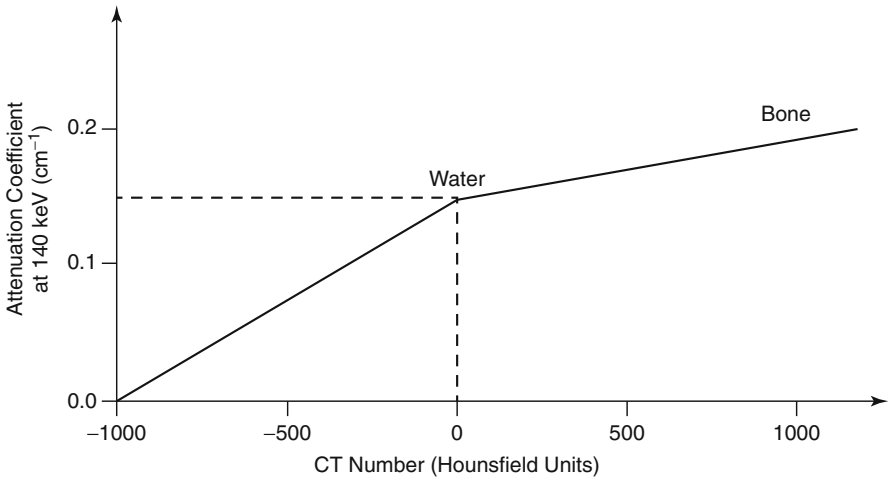

Fig. 4.10 The relationship between attenuation coefficient and CT number for 140 keV gamma rays

create an attenuation map, these numbers need to be converted into attenuation coefficients at the required energy. This is usually done using a bilinear graph like the one shown in Fig. 4.10. Air has a CT number of −1,000 HU and an attenuation coefficient of 0.0 cm^{-1}. Water has a CT number of 0 HU and an attenuation coefficient of 0.15 cm^{-1} at 140 keV. Bone has a CT number of 500 HU or more and a slightly higher attenuation coefficient. So for every pixel in the CT image, this graph can be used to convert its CT number into the corresponding attenuation coefficient and thus create an attenuation map. The graph shown in Fig. 4.10 is the one that would be used for an emission energy of 140 keV, but similar graphs can be created for any other desired emission energy.

The main drawback of using a CT scan to create the attenuation map is that it has to be acquired immediately before or after the emission scan rather than simultaneously as was the case with ^{153}Gd. This creates two potential problems. Firstly, there is an opportunity for the patient to move between the two scans, so care must be taken to check the two scans for proper registration (see Sect. 4.5.5). Secondly, the duration of the emission scan will be several minutes, whereas the duration of the CT scan might only be a few seconds. This can cause additional registration problems due to respiration. The emission scan will inevitably be averaged over many breathing cycles, but the CT image could be taken at full inspiration or full expiration. This can lead to a mismatch between the position of the diaphragm on the emission image and on the attenuation map, which in turn can lead to incorrect attenuation correction. Some manufacturers have addressed this problem by producing CT systems that deliberately acquire slow CT images so that they have the same respiration averaging as the emission image. The features of different hybrid SPECT/CT systems will be compared in Chap. 6.

4.4.3 Scatter Correction

Unlike attenuation, which results in a loss of gamma rays, scatter results in a gain of gamma rays but in the wrong place. In Fig. 4.9a, gamma ray "A" was a good event which did not interact in the patient and was absorbed in the crystal by photoelectric absorption, giving up all its energy (140 keV for ^{99m}Tc), so in the energy spectrum, Fig. 4.9b, it contributes to the photopeak events. Because of the poor energy resolution of the gamma camera crystal, although these events really all have an energy of exactly 140 keV, they appear spread out in energy, and so we usually allow an energy window of ±10 % to accept these good events. Now gamma ray "D" was heading in the wrong direction and should not have been detected at all, but after undergoing Compton scatter in the patient, it travels in a direction that allows it to get through the collimator. Since it has been scattered, it will have lost some energy, and one would hope that it would be rejected by the energy window. However, it turns out that a scatter angle of 45° only reduces the gamma ray energy from 140 to 130 keV, which is still within the usual energy window of 140 ± 14 keV. Therefore, gamma rays scattered by up to 45° will still be accepted in the image. But because of the scatter, these gamma rays appear to have come from part of the patient where there is no activity – shown by the dashed line in Fig. 4.9a. This means that image contrast, the difference between normal and abnormal features, is reduced by scatter. Scatter correction is designed to correct for these scattered events and hence improve image contrast.

Scatter correction is usually performed by acquiring one or more additional images using energy windows that are set to sample the scattered events. Two common methods are those developed by Jaszczak [11] which uses a single scatter window below the photopeak and Ichihara [12] which uses two scatter windows at either side of the photopeak. In both methods, the scatter windows are assumed to represent the distribution of scattered radiation only, and so scatter correction is performed by subtracting a fraction of the scatter image from the photopeak image.

4.4.4 Iterative Reconstruction

Iterative reconstruction is an entirely different approach to reconstruction of SPECT images, and it offers an alternative to filtered back projection. Iterative reconstruction is not actually a new technique. It goes back just as far as filtered back projection, but because it requires a lot more computation, it was always considered too slow for routine clinical work. However, with the development of improved algorithms that have speeded up the calculations, together with the increase in speed of modern computers, iterative reconstruction is now a viable method, and it is available from all the major manufacturers.

Iterative reconstruction works on the principle of first guessing what the answer should be and then repeatedly refining the guess until it agrees with the data. The process is illustrated in Fig. 4.8b. The first step is to make an initial estimate of what

the transaxial activity distribution might be. A common starting point is to assume that all the counts from the acquired projection images are uniformly spread across the transaxial image. The next step is to calculate what the projection images would look like if this estimated transaxial distribution was correct. This step involves a forward projection of the transaxial counts into a set of estimated projection images. Of course these estimated projections will not be the same as the real acquired projections, but by comparing the estimated and acquired projections, the error can be calculated. If this error is back projected, a correction term can be deduced which is used to adjust the estimated transaxial image. The process of forward projection, comparison, back projection, and adjustment is then repeated many times, and each time round the loop is called an iteration. Hopefully, after sufficient iterations, the estimated projections become very similar to the real acquired projections at which point the corrections make negligible difference and the process is said to have converged. At this point, if the estimated projections are the same as the acquired projections, then one can assume that the estimated transaxial image is the same as the real transaxial image, and so the reconstruction has achieved its objective.

Early iterative methods developed in the 1970s, such as the algebraic reconstruction technique [13] and the simultaneous iterative reconstruction technique [14], were very slow to converge and required hundreds of iterations, so they would take hours of computing on the slow computers of the day. A big step forward came in 1982 when Shepp and Vardi developed the MLEM (Maximum Likelihood Expectation-Maximization) algorithm [15]. Although this was still slow, it had the advantage that it took account of random noise in the projection images. Then in 1994, Hudson and Larkin produced a modification of this algorithm called OSEM (Ordered Subsets Expectation-Maximization) [16]. This algorithm converges to a solution much more quickly and so requires only a few iterations.

OSEM is therefore the basis of most commercial implementations of iterative reconstruction. It speeds up the convergence by splitting the acquired projections into several subsets, so, for example, 60 projections might be grouped into 10 subsets with 6 projections in each subset. The current estimate of the transaxial image is then updated after every subset has been compared, rather than waiting until all projections have been processed. To use the OSEM algorithm, the user only has to specify the number of subsets (10 in the above example) and the number of iterations (e.g., 2). The time taken depends on the number of iterations, but the quality of the image depends on the number of updates, which is equal to the number of iterations multiplied by the number of subsets. Thus, MLEM with 20 iterations (and implicitly only 1 subset because MLEM does not use subsets) would give the same image quality as OSEM with 2 iterations and 10 subsets, but the OSEM would be ten times faster.

Now the accuracy of the result using any iterative reconstruction algorithm depends on the accuracy of the forward projection step in Fig. 4.8b. The accuracy of the back-projection step is not so important since as the process converges, the error that is to be back projected tends toward zero anyway. The forward projection step is essentially a modeling of the acquisition process from an estimated activity distribution in the patient to an estimated projection. So if the modeling is completely

realistic and takes account of real processes such as attenuation, scatter, and loss of resolution with distance, then the final answer will be an estimate of the real activity distribution that takes account of these effects. Thus, a major benefit of iterative reconstruction is that it can incorporate attenuation correction and scatter correction into the reconstruction process rather than treating them as add-ons as with filtered back projection.

If an attenuation map has been created, for example, using a hybrid SPECT/CT system, then it is easy to incorporate this into the forward projection step. Instead of just adding up counts from all pixels in the estimated transaxial that lie along the projection ray, counts from each pixel are reduced by an appropriate amount corresponding to attenuation from all pixels in the attenuation map along the projection ray. This gives an estimate of what the projection image would be, including the calculated effect of attenuation. This is compared with the acquired projection image, which includes real attenuation. So like is compared with like, and attenuation correction will automatically be achieved. This is why iterative reconstruction has become the standard method whenever nonuniform attenuation correction is required.

So far the assumption has been that iterative reconstruction is performed for one transaxial slice at a time in the same way that it is for filtered back projection. This is known as 2D iterative reconstruction because each 2D slice is reconstructed before moving on to the next slice to build up the full 3D data set. But there is an alternative method known as 3D iterative reconstruction where all slices are reconstructed simultaneously. This requires that the forward projection step models the gamma camera acquisition process in full 3D rather than just one slice at a time. Full 3D iterative reconstruction involves more calculations and also tends to need more iterations, so it is slower than 2D iterative reconstruction. Nevertheless, it is now practical with modern computers, and it is offered as an option by most manufacturers.

Attenuation of gamma rays occurs only along the projection ray, and so it only affects one slice at a time. Therefore, attenuation correction is dealt with adequately by 2D iterative reconstruction. However, scatter of gamma rays can occur in all directions, and so scatter is affected by activity in other slices. Therefore, if scatter is to be modeled within the forward projection step, this must be done using 3D iterative reconstruction, and some manufacturers have used this approach. However, others have stuck with the traditional method of measuring scatter using additional energy windows (see Sect. 4.4.3). In this method, they perform forward projection without worrying about scatter and then add in a measured scatter contribution from the scatter window. This method can be applied in either 2D or 3D iterative reconstruction.

4.4.5 Resolution Recovery

For many years, a fundamental limitation of nuclear medicine images has been its poor resolution, but now 3D iterative reconstruction offers the possibility of doing something about this. If the blurring of the image by the collimator is modeled within the forward projection step, then the resulting estimated blurred projection

image can be compared with the real projection image that includes real collimator blurring. Once again like is compared with like, and so the final result should be a reconstructed image that takes account of collimator blurring. This is referred to as *resolution recovery* because the reconstructed image should have better resolution than the acquired projection images, truly a benefit worth having. All that is necessary to achieve this is to know the system resolution of the collimator that is used (using a graph like Fig. 4.4) and the distance of the camera from the patient. The former is easy to measure or calculate from a knowledge of the size of the collimator holes. The latter can be recorded by the camera software during the acquisition or estimated from orthogonal views. However, since collimator resolution blurs the image between slices as well as within slices, resolution recovery definitely requires 3D iterative reconstruction, not just 2D.

4.4.6 Dealing with Noise

If iterative reconstruction software can include attenuation correction, scatter correction, and resolution recovery, then it sounds as if it ought to be able to produce perfect answers. However, there is still one limiting factor, which is image noise. Noise is a result of random fluctuations inherent in the radioactive decay process itself, and so there is no way of avoiding this.

We have already seen how filtered back projection has the unfortunate effect of amplifying noise, and so it is essential to incorporate a smoothing filter to reduce noise, and this will also spoil the resolution. Fortunately, iterative reconstruction using either MLEM or OSEM does not amplify noise in the same way, because both of these algorithms take account of noise in the projection images. However, if too many iterations are used, then noise does begin to appear in the reconstructed sections, and so it is normal to stop the process after a few iterations. Therefore, the number of iterations becomes the parameter that controls the balance between image smoothness and noise. Most iterative reconstruction software also allows the user to apply a final smooth to the resulting image to remove residual noise. Unlike filtered back projection, the smooth in iterative reconstruction is not an essential part of the process; it is just an optional cosmetic smooth to adjust the final image to the user's preferences.

Now there is always a desire to get away with acquiring as few counts as possible, but as image counts reduce, then noise gets worse, and so there is a practical limit to the minimum number of counts required for a sensible image. Of course counts can always be increased by increasing administered activity or by increasing acquisition time, but those have associated disadvantages of increased radiation dose or reduced patient comfort. Actual counts can also be increased by using a collimator with bigger holes which gives improved sensitivity, but the disadvantage of this is that resolution will be worse. However, with the availability of resolution recovery software, it is now possible to model the poor collimator resolution and correct for this. In fact, some novel designs of SPECT system make use of this

approach to obtain scans in a very short time using high-sensitivity but low-resolution collimators and then reconstruct using resolution recovery software to give a good quality image.

Using conventional cameras, there are still software solutions to dealing with studies that have low counts which would normally be too noisy even for standard OSEM software. The latest versions of 3D iterative reconstruction software also incorporate techniques for dealing with very low counts. For example, the MAP (Maximum A Posteriori) algorithm [17, 18] modifies the OSEM algorithm to allow for noise in the reconstructed images rather than noise in the projection images. By incorporating prior information, for example, that there should not be sudden changes between adjacent pixels, it can favor smooth solutions over noisy ones. Manufacturers now offer reconstruction software that can produce satisfactory results from studies with half or even one-quarter of the normal counts. This offers the option to either acquire shorter studies or to use less activity without compromising image quality.

4.5 Quality Control

Quality control is important for all gamma camera studies, but it is particularly important for SPECT/CT because problems can either cause image artifacts which may be mistaken for abnormalities or create poor images where abnormalities may be missed.

4.5.1 Uniformity

All gamma cameras should be checked regularly to ensure that they produce uniform images, but this is even more important if the camera is used for SPECT. The reason for this is that the reconstruction process amplifies any nonuniformities that occur near to the axis of rotation. A pixel near to the edge of the camera field of view will image a different part of the patient in each projection, and so any defect at this point will be averaged out with other good pixels. However, any pixel close to a line down the center of the field of view will image the same part of the patient in all projections, and so any defect at this point will be greatly amplified. The effect of this is that if the camera has any areas of poor uniformity (either hot or cold) close to its center line, then the reconstructed transaxial images will show corresponding hot or cold rings around the axis of rotation. The effect becomes less pronounced further away from the axis of rotation, but the appearance will change from slice to slice because each slice is imaged by a different part of the camera field of view.

In order to avoid these artifacts, only cameras in good working order with all energy and linearity corrections applied should be used for SPECT. In addition, a final sensitivity correction should also be applied to the projection images. The

camera uniformity should be monitored weekly using planar uniformity images, and any change should be compensated by acquisition of a new sensitivity map. The sensitivity correction map needs to be acquired as a high-count flood image containing 100 million counts in order to give adequate statistical precision, preferably using the appropriate collimator, as collimator defects are a likely cause of such problems.

4.5.2 Center of Rotation

The center of rotation (COR) is the point in the camera's field of view where a point source placed on the axis of rotation would be imaged. Since the camera rotates about this fixed axis, the COR will remain at a fixed position in the camera's field of view. SPECT reconstruction software needs to know where the COR is in order to perform the back projection properly, and by default, it assumes that the COR is exactly in the middle of each projection image. Normally the COR will actually be quite close to the center of the image, but if there are mechanical or electronic misalignments, it may be slightly offset. Therefore, manufacturers always provide a means to measure this COR offset and to record it with the acquired data. If the COR offset is wrong, then back-projection rays will not converge to a single point and the reconstructed images will be blurred. Therefore, it is advisable to check that the COR offset has not drifted on a monthly basis. The check should be done with all collimators used for SPECT and done separately for H-mode and L-mode if both are used clinically.

The COR offset can easily be checked by placing a small volume source somewhere near to the center of the field of view, 0.1 ml of ^{99m}Tc solution in the end of a 1 ml syringe will do. Then the system is set to acquire a SPECT study using a full 360° orbit. If the system has two detectors, they should be analyzed separately, and so each detector must be set to acquire a full 360° orbit rather than letting each acquire just one-half of the orbit as would be usual for a clinical study. If by chance the source happens to have been placed exactly on the axis of rotation, then it will appear to remain stationary on the projection images and its X-coordinate will define where the COR is, usually though the source will not be on the axis of rotation, and so it will appear to wobble about the COR as seen on the projection images. Therefore, a graph of the X-coordinate of the position of the center of the source plotted against gantry angle will show a sinusoidal variation about the COR. A sinusoidal fit to the data can be used to determine the midpoint, and if this is compared to the center of the image matrix, this will be the COR offset. This should correspond to the COR offset stored by the system to within 1 mm.

At the same time, it is helpful to determine the Y-coordinate of the center of the point source, because as long as the detector heads are level, this should not show any sinusoidal variation. The absolute value of the average Y-coordinate of the source position is arbitrary, because it just reflects where the source was placed. However, for a double-headed system, the mean Y-coordinate should be the same for the two detectors. If it is not the same, this represents a misalignment between the two detectors which can also lead to reconstruction artifacts.

4.5.3 Resolution

SPECT resolution can be checked with a phantom containing several fine capillary tubes filled with ^{99m}Tc solution. Some of the tubes are placed parallel to the axis of rotation and some perpendicular to it. A SPECT acquisition is performed using standard clinical parameters but with an increased acquisition time to give plenty of counts. The image should be reconstructed without any smoothing filters in order to give the best possible resolution. Then profiles taken across the reconstructed images of the tubes can be used to determine the resolution from the full width at half maximum of the profiles. This test should be performed to give a baseline when the system is first installed, but thereafter, it only needs checking annually.

4.5.4 Patient Movement

If the patient moves during the SPECT acquisition, this can spoil the reconstruction and can produce artifacts or make the image uninterpretable. Therefore, every scan should be checked to ensure that there is no patient movement before the patient is allowed to leave. Small patient movements may be correctable by motion correction software, but if there is any significant movement, the acquisition should be repeated.

If the patient moves along the axis of rotation, this can easily be seen as an up and down motion on the projection images if they are displayed as a cine sequence. However, sideways movement is not easily seen on the cine display because the image naturally oscillates sinusoidally anyway. The sinogram display is an easier way to detect sideways motion. The *sinogram* is an image made up by taking one row from each projection image and stacking them together. Therefore, the horizontal distance on the sinogram represents position across the projection image, and vertical distance up the sinogram represents angle around the patient. A normal sinogram should therefore show bits of the patient describing smooth sinusoidal curves. Any sudden discontinuities in the sinogram are an indication that the patient moved sideways. On a dual-headed system, a discontinuity halfway up represents either a misalignment between the two detectors or slow patient creep during the study.

4.5.5 Registration Between SPECT and CT Images

Another important part of quality control that is unique to hybrid SPECT/CT systems is to verify that the SPECT and CT images are correctly registered. The main advantage of having a hybrid system where the patient does not have to move off the bed between acquisition of the SPECT and CT images is that this should ensure proper alignment between the two images. However, things can go wrong, and if they do, this can cause artifacts in the resulting images.

In most SPECT/CT systems, the patient bed has to advance by a known distance between the SPECT and CT systems, and this will be calibrated by the engineers using a special phantom at installation time. Even if this special phantom is not available, the user can easily check that the registration is correct by imaging any SPECT phantom filled with ^{99m}Tc. If the phantom has parallel rods, like the Jaszczak phantom, then it is best to place it at an angle to the axis of the bed otherwise longitudinal positioning errors may not be obvious. If a SPECT/CT study is acquired and the fused SPECT and CT images are displayed with standard clinical software, then any mismatch between the radioactive distribution and the plastic of the phantom should be visible.

One difference between a phantom and real patient studies is that the phantom will not be as heavy as a patient, and it is possible that the bed may bend under the weight of a patient. Therefore, it is a good idea to put extra weights on the bed to simulate the weight of a patient when performing this test. If the system has a cantilevered bed, then rollers are usually provided to support the far end of the bed and prevent it from sagging. However, if the bed is not touching the rollers for any reason, then some sagging will occur and misregistration can result.

The other difference between phantoms and patients is that patients can move, so even if registration is perfect with a phantom, it is still important to check it with every patient study. If the radiopharmaceutical has sufficient uptake in normal tissues, then a fused SPECT/CT display should show any gross misalignment of the two images. Some manufacturers provide software that makes this easier by showing the SPECT image as a contour superimposed on the CT image. This is particularly helpful for myocardial perfusion SPECT where there is a sharp edge on the CT between heart and lung, and the contour of the myocardium on the SPECT image should align with this. Sometimes the software can be used to make small adjustments to correct minor misregistration. If the errors are large, then the study may need to be repeated, but since this will involve a repeat of the CT study, the additional patient dose will need to be considered.

If a misregistration of the SPECT and CT images is not corrected, then the consequences for a study used for anatomical localization are obvious; the SPECT abnormality will not appear in quite the right place when overlaid on the CT. Whether or not this has any diagnostic consequences will depend on the type of study and the magnitude of the error, but it could easily make a solitary hot spot appear in soft tissue instead of in bone, for example.

If the CT study is used only for attenuation correction, then the consequences of misregistration are not so obvious but can be equally misleading. If the attenuation map is shifted relative to the SPECT image, then the wrong value of attenuation correction will be applied and some pixels may be scaled up by too much or too little. For example, in myocardial SPECT, the boundary between heart and lung tissues involves a big step in attenuation coefficient, so a small shift of the attenuation map toward the heart can cause some pixels in the myocardium to be under-corrected. This has been reported to create apparent perfusion defects in the anterior myocardial wall which disappear when the misregistration is corrected [19].

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Chapter 5
Computed Tomography: Physical Principles, Instrumentation, and Quality Control

Euclid Seeram and Joanne Sil

5.1 Introduction

CT scanning has advanced rapidly over the last few years to become one of the most used diagnostic imaging tools available. When CT scanning was introduced in the early 1970s, data was acquired incrementally in a *step and shoot* manner; one image was produced per tube rotation while the patient and couch remained stationary; the patient and couch would then be moved by a predetermined length and the process would be repeated. Needless to say, this was an extremely lengthy process, and the images produced (while revolutionary at the time) were not comparable in quality to the images produced by today's modern scanners.

Since then, CT technology has evolved through various generations of scanners to give first, single-slice helical scanners and then the multi-slice CT scanners that are currently in operation today. The availability, versatility, and efficiency of contemporary diagnostic CT scanners together with excellent image quality available have made it a favorable imaging modality for a wide range of clinical situations. In addition to CT being an established modality in its own right, it is now highly utilized in hybrid imaging. However, the acquisition of CT data requires a higher radiation dose than more traditional diagnostic X-ray imaging procedures. This coupled with the increased utilization of both diagnostic CT and low-dose CT (as used in SPECT-CT) makes it imperative that quality procedures are put in place to retain

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adequate and consistent performance of the equipment, to guarantee patient safety, and to ensure optimal radiation protection.

The purpose of this chapter is to outline the physical principles of CT, describe the major elements of the technology (hardware components) and basic quality control principles for the CT scanner, as well as highlight several technical applications, including applications in cardiac imaging, fusion imaging, and radiation treatment planning.

In the early 1970s, Dr. Godfrey Hounsfield (in England) invented the first clinically useful CT scanner using an improved reconstruction algorithm. For this work, Hounsfield received the Nobel Prize in Medicine and Physiology in 1979, a prize which he shared with Allan Cormack, a physicist at Tufts University in Massachusetts. Dr. Hounsfield's CT scanner was called the EMI (Electronic and Musical Instruments) scanner since this was the company where he worked. The EMI scanner was dedicated to imaging the head only.

Professor Cormack developed solutions to the mathematical problems in CT. Later in 1963 and 1964, he published two papers in the *Journal of Applied Physics* on the subject, but they received little interest in the scientific community at that time. It was not until Hounsfield began work on the development of the first practical CT scanner that Dr. Cormack's work was viewed as the solution to the mathematical problem in CT.

The earlier scanners were referred to as conventional slice-by-slice CT scanners. These scanners took a long time for imaging patients. It then became a challenge to increase the volume coverage speed without compromising image quality. In this regard, another significant contribution to the development of CT technology came from work by others, most notably Dr. Willi Kalender, a physicist in Germany, who introduced the notion of scanning the patient continuously to cover a volume of tissue instead of a single slice. These scanners became known as single-slice spiral/helical CT scanners or volume CT scanners since their volume coverage speed was much faster than conventional single-slice CT scanners.

Later, in 1998, a new generation of CT scanners was introduced at the Radiological Society of North America (RSNA) meeting in Chicago. These scanners are called multi-slice volume CT scanners because they were based on the use of multi-detector technology to scan 4, 8, and 16 and 32, 40, 64, and 320 slices per gantry rotation, thus increasing the volume coverage speed compared to single-slice volume CT scanners.

Multi-slice CT (MSCT) is now state of the art in CT scanning, and it offers a wide range of applications such as CT fluoroscopy, CT angiography, three-dimensional imaging, and virtual reality imaging. Additionally, MSCT scanners now play an active role in fusion imaging as well as in radiation treatment planning.

5.2 The CT Process: Major System Components

There are three major steps in the production of a CT image – as is illustrated in Fig. 5.1. These include data acquisition, image reconstruction, and image display, storage, and communication. Each of these will now be described.

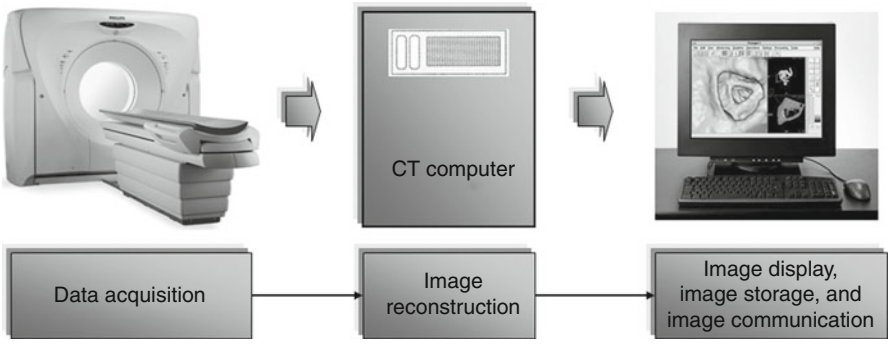


Fig. 5.1 The three major steps in the production of a computed tomography (*CT*) image: data acquisition; image reconstruction; and image display, recording, storage, and communication. See text for further explanation

Data acquisition is characterized by scanning the patient systematically to collect X-ray attenuation data (sometimes referred to as attenuation readings) that are measured by an array of special electronic detectors, coupled to the X-ray tube. This coupling defines a third-generation beam geometry that has become the most commonplace geometry (size and shape of the X-ray beam) used in CT systems. During scanning, the X-ray tube and detectors rotate around the patient to collect X-ray transmission measurements.

There are two types of data acquisition methods. These include slice-by-slice data acquisition typical of conventional *stop-and-go* CT scanners and the more current approach – volume data acquisition that is typical of MSCT scanners. In the former, the X-ray tube and detectors rotate around the patient to collect attenuation readings for the first slice. The tube then stops, and the patient is prepared for the second slice. This process continues until all individual slices have been collected from the volume of anatomy of interest. For volume data acquisition, the X-ray tube rotates continuously around the patient; during this process, the patient moves through the CT gantry at the same time to cover the entire volume of anatomy under investigation.

Image reconstruction involves the use of the attenuation readings sent to the computer for processing. The major task of the computer is to reconstruct a digital image using algorithms to generate the image. One of the more commonplace reconstruction algorithms used in CT is filtered back projection; this will be described later in this chapter.

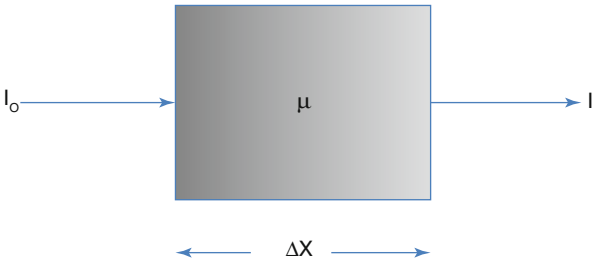
Finally, after the image has been reconstructed, it is displayed on a computer screen for viewing by an observer. At this point, the observer can manipulate the image using image processing software to suit their viewing preferences and to best demonstrate pathology and anatomy. Images are then stored on magnetic or optical data carriers, and they can be communicated by electronic means to other locations. Electronic communications in CT require a standard protocol that facilitates connectivity (networking) among imaging modalities and multi-vendor equipment. The standard used for this purpose is the Digital Imaging and Communications in Medicine (DICOM).

5.3 Basic Physical Principles of CT

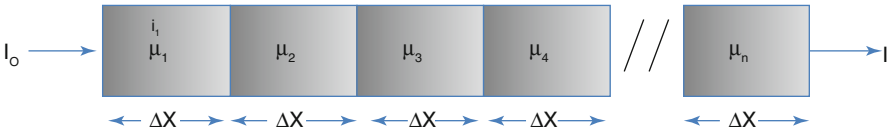
In CT as the beam passes through the patient, it is attenuated according to Lambert-Beer's Law, expressed as

$$I = I_0 e^{-\mu \Delta x}$$

where I = transmitted beam intensity, I_o = original beam intensity, e = Euler's constant, μ = linear attenuation coefficient, and Δx = finite thickness of the section. Graphically, this can be shown as



As the beam passes through a stack of voxels (volume elements) that is part of the CT slice, an attenuation measurement called a ray sum is obtained. A ray sum is the sum of all μs along the path of a single ray through the patient as shown below:



In this situation, the transmitted intensity, I, is represented as

$$I = I_0 e^{-\sum_{i=1}^{n} \mu_i \Delta x}$$

or

$$-\sum_{i=1}^{n} \mu_i \Delta x = -(\mu_1 + \mu_2 + \mu_3 + \ldots \mu_n)\, \Delta x$$

By taking the natural logarithm (ln), this equation becomes

$$\ln\left(I_0 / I\right) = \sum_{i=1}^{n} \mu_i \Delta x$$

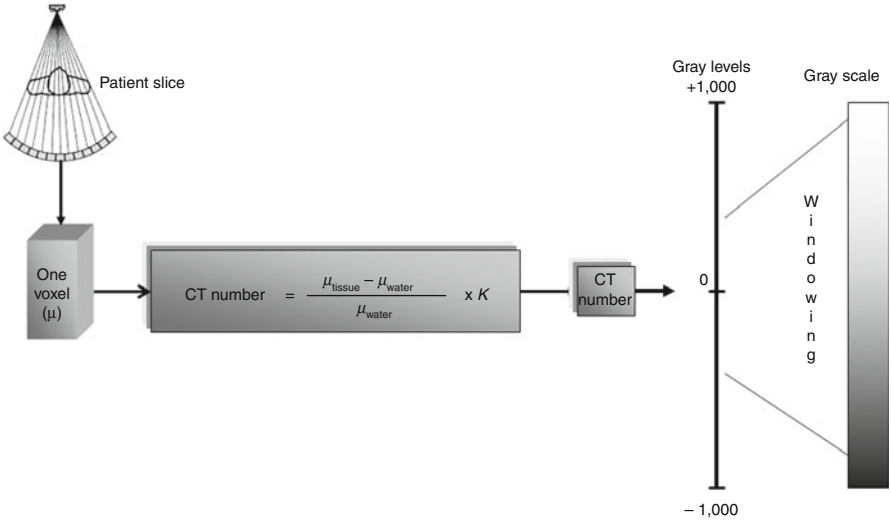


Fig. 5.2 The relationship between the attenuation coefficient (μ) in a tissue voxel and the computed tomography (*CT*) number

The problem in CT is to calculate each of the μs in the entire slice. Therefore, the image reconstruction process requires that a large number of ray sums (set of transmission measurements) be obtained for different locations (rotation angles) around the slice to be imaged. In reality, the slice to be imaged may consist of a 512 × 512 matrix or 1,024 × 1,024 matrix made up of voxels. Each voxel attenuation data (μ) is converted into a CT number (an integer), forming a matrix of numbers. The digital matrix is the digital image that has been created using a specific image reconstruction algorithm. Figure 5.2 shows how the attenuation data are converted to integers referred to as CT numbers using a reconstruction algorithm.

These CT numbers are computed using the following relationship:

$$\text{CT Number} = \frac{\mu_{\text{tissue}} - \mu_{\text{water}}}{\mu_{\text{water}}} \cdot K$$

where K is a manufacturer's scaling factor or contrast factor and, in general, $K = 1{,}000$.

The relationship between tissue voxel μ and image pixel (CT number) is shown in Fig. 5.2. After these calculations, a matrix of CT numbers is generated in the computer. These numbers can be printed out as a numerical image; however, since radiologists prefer to view grayscale images, the numbers (digital matrix) must be converted into a grayscale image. This conversion is illustrated in Fig. 5.3. The following points are noteworthy: First, the digital image is a matrix of CT numbers which are the CT numbers converted into shades of gray where the higher numbers are assigned white, lower numbers are black, and gray shades are between black and

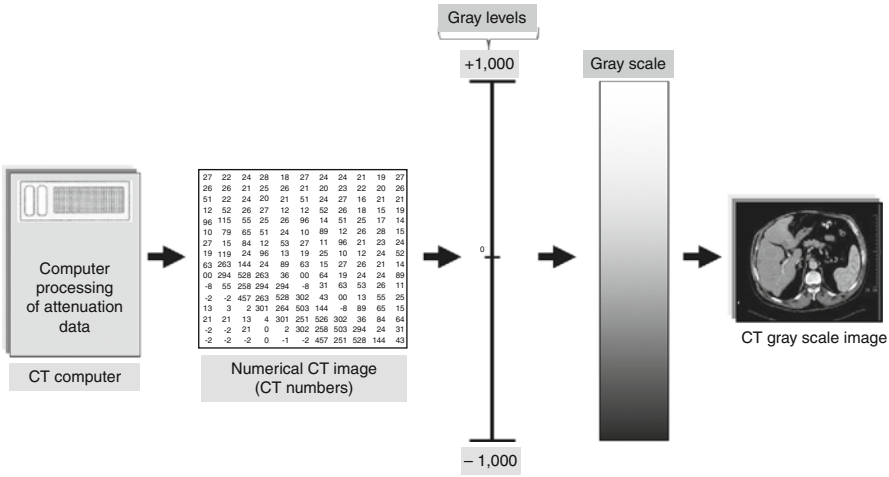


Fig. 5.3 The conversion of computed tomography (*CT*) numbers into a grayscale image. See text for further explanation

white. This assignment is related to the attenuation characteristics of tissues. Bone attenuates more radiation and therefore is assigned white (the same appearance as an X-ray film-screen image) while air attenuates very little radiation and appears black (as on an X-ray film-screen image). Secondly, the range of CT numbers is defined as the window width (WW). The center of the range is defined as the window level (WL), and finally, by manipulating the WW and WL, the image contrast and the image brightness can be altered, respectively. This operation is called windowing, and it will be described later in this chapter.

5.3.1 Image Reconstruction Algorithms

It is not within the scope of this chapter to describe the details of reconstruction algorithms in CT; however, the basic ideas of image buildup using a set of transmission measurements taken from a large number of positions around the patient are in order. It is important to note that since the essential task of these algorithms is based on calculating the image (so to speak), a computer is a central component of the CT scanner.

A more complete definition of this technique has been given by [1] who states that "image reconstruction from projections is the process of producing an image of a two-dimensional distribution (usually of some physical property) from estimates of its line integrals along a finite number of lines of known locations." In radiology, the physical property is the attenuation coefficients of the tissues, while the line integral refers to the sum of the attenuation along each ray in the X-ray beam that passes through the slice of the patient. The number of lines of known locations refers to the various positions of the X-ray tube and detectors as they rotate around the patient.

Image reconstruction from projections has its theoretic roots in 1917 when Radon, a mathematician, applied these techniques to solving gravitational problems. Later, these image reconstruction techniques were applied to solving problems in astronomy and optics and then, subsequently, in medicine. It was in 1971 that others such as Oldendorf, Kuhl, and Edwards produced images using reconstruction techniques. Unfortunately, their images were blurred and served no useful purpose in diagnostic radiology. This problem was overcome by Godfrey Hounsfield in England, who used algebra to solve a large number of simultaneous equations that were generated from the data collected by the X-ray tube and detectors as they rotate around the patient. The use of a computer proved beneficial to solve these equations; however, the computer took some time to perform these calculations, and therefore more efficient methods were needed.

The systematic set of rules for solving a problem is referred to as an algorithm, and Hounsfield's algorithm was referred to as the algebraic reconstruction technique. While the earlier reconstruction algorithm was called back-projection algorithm and produced image blur, more efficient algorithms called Analytic Reconstruction Techniques (ART) were developed. One such popular ART algorithm used in CT scanners in the past was the filtered back-projection algorithm. This algorithm gets rid of the blur produced when the back-projection algorithm is used.

The essential steps of the back-projection and the filtered back-projection algorithms are shown in Fig. 5.4. In back projection, the projection profiles are collected and then back projected (i.e. linearly smeared as shown in Fig. 5.4 to form the image). The star pattern that appears results in image blur, and the image is not useful to a radiologist. Getting rid of the classical star pattern (blur) requires the use of the filtered back-projection algorithm. In Fig. 5.4, each profile collected is filtered using a digital filter known as convolution. The filtered projections are summed, and the negative and positive components are canceled, and this produces an image free of blurring.

MSCT scanners now use different algorithms that are necessary to deal with the fact that in these scanners the patient is moving as the X-ray tube and detectors rotate continuously. Interpolation algorithms were developed to first produce projections in a single plane (planar section) and subsequently coupled with the filtered back-projection algorithm to produce images free of artifacts as a result of continuous patient movement during the scanning process. As CT scanners continued their evolution beyond 16 slices per revolution, other algorithms referred to as "cone beam" algorithms were developed – these are now commonplace. It is not within the scope of this chapter to describe the details of these algorithms, since they are complex and are not necessarily in the practical domain of the CT imaging operator.

5.3.2 CT Instrumentation: Basic System Components

The basic equipment configuration for a CT scanner is shown in Fig. 5.5. Three major components are clearly apparent: the imaging system, the computer system, and the image display, recording and storage, and communication systems. The

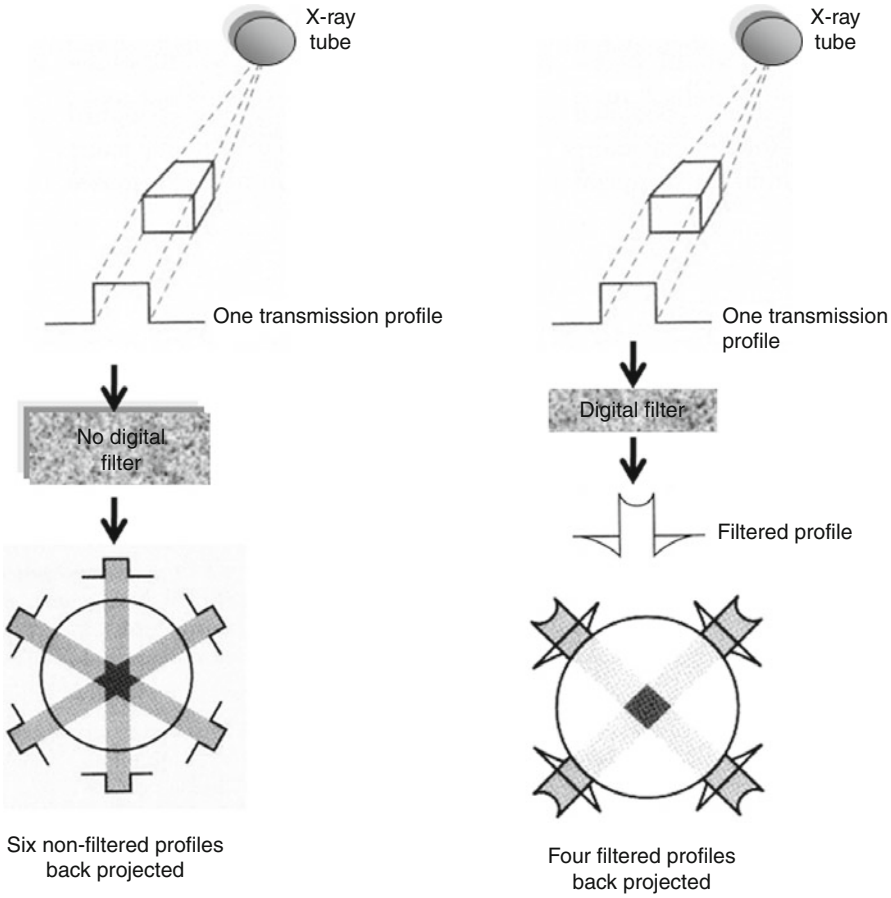


Fig. 5.4 The essential steps of two types of early image reconstruction algorithms: the back-projection algorithm (**a**) and the filtered back-projection algorithm (**b**). The purpose of the filtered back-projection algorithm is to remove the classical blur characteristic of the back-projection algorithm

imaging system consists of equipment that collectively is referred to as the data acquisition components.

Data acquisition is a term used to describe the systematic collection of X-ray transmission readings (attenuation data) from the patient. Data acquisition components for CT include the X-ray generator and X-ray tube, a beam-shaping filter, pre-patient collimator, (patient in the gantry) electronic detectors, and the analog-to-digital converters (ADCs) as is illustrated in Fig. 5.6.

The manner in which the X-ray tube and detectors are arranged as well as the scanning motion used to collect attenuation data from the patient is referred to as the data acquisition geometry. The term geometry refers to the size, shape, and motion and the path traced by the X-ray tube. Today, at least six generations of CT scanners have emerged, and their categorization is based on the scanning geometry, scanning motion, and the number of detectors used in the scanner. The overall goal of this evolution is to improve the data collection speed while maintaining image quality.

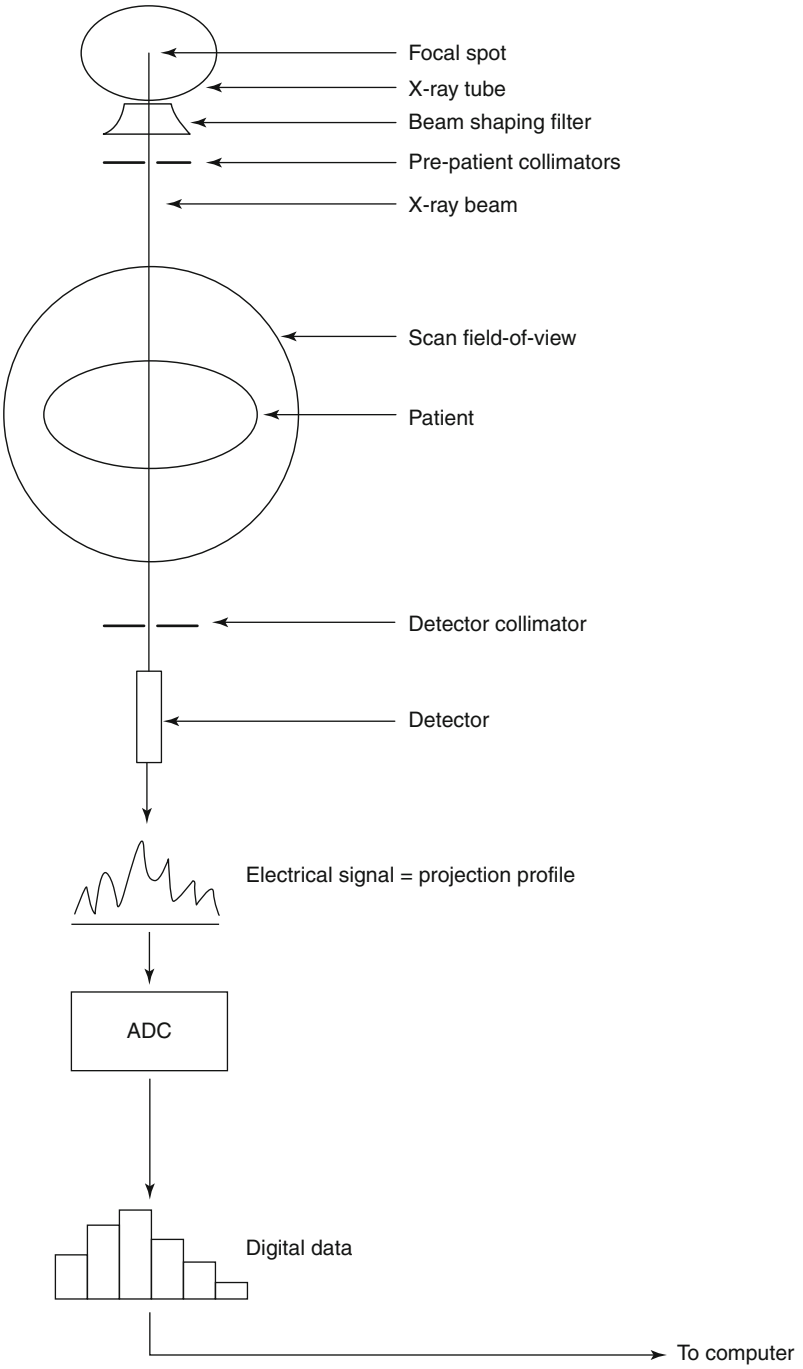


Fig. 5.5 The major components of the data acquisition system of a computed tomography scanner. These include the X-ray tube, filter, collimator, patient and the scan field of view, detector, and the detector electronics that includes the analog-to-digital converter (Reproduced, by permission, from Seeram [2]. Copyright © 2009 by Saunders, an imprint of Elsevier Inc.)

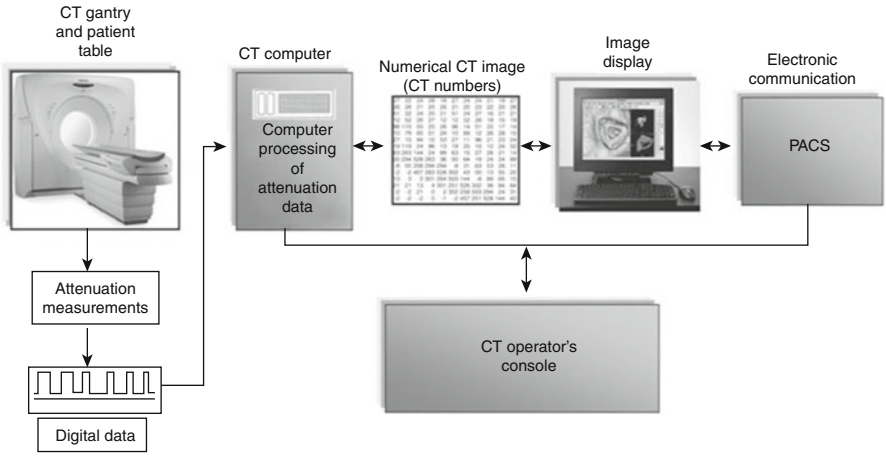


Fig. 5.6 The computed tomography (*CT*) scanner gantry (which contains all the components in Figs. 5.4 and 5.5) sends all data signals to the computer for image reconstruction. The results of computer processing are images that are displayed on a television monitor for viewing by an observer. PACS, picture archiving and communication system (Reproduced, by permission, from Seeram [2]. Copyright © 2009 by Saunders, an imprint of Elsevier Inc.)

In Fig. 5.6, the scanner gantry contains all the data acquisition components shown in Fig. 5.5. The tube and detectors rotate around the patient to collect a large number of attenuation readings. The X-ray beam is shaped by a special filter called a bow-tie filter. This specially shaped filter is intended to make the beam more uniform at the detector (since a heterogeneous beam is used in CT to allow for the use of Lambert-Beer's law to be used to compute μ, this law only holds true for a homogeneous beam of radiation which Hounsfield used in his original experiments). In addition, the beam is collimated to pass through only the slice of interest. It is attenuated by the patient, and the transmitted radiation is measured by the detectors that convert X-rays into electrical signals. These signals converted into digital data are subsequently sent to the computer for image reconstruction.

CT detectors fall into two categories – scintillation detectors and gas-ionization detectors. Scintillation detectors (solid-state detectors) use a scintillation crystal (e.g., cadmium tungstate) coupled to a photodiode. The crystal captures the radiation and converts it into light. The light falls upon the photodiode that converts it into an electrical signal. This signal is sent to the ADC for digitization. Gas-ionization detectors on the other hand are chambers filled with xenon gas. X-rays passing through the patient fall upon the volume of gas, and ionization results. The electrical signal arises from the ionization and is digitized by the ADC. Gas-ionization detectors have become obsolete and are not used in today's MSCT scanners.

An important point to note here relates to the ADC. The ADC divides the electrical signal into multiple parts – the more parts, the more accurate the ADC. The parts are measured in bits: A 1-bit ADC divides the signal (samples) into two digital values (2^1), a 2-bit ADC generates four digital values (2^2), and a 12-bit ADC generates

4,096 (2^{12}) digital values. These values help to determine the grayscale resolution of the CT image. Modern CT scanners use 16-bit ADCs.

Essentially, the data collected from the patient flows from data acquisition devices, including ADCs, to the preprocessor and subsequently to the host computer. The data is stored as raw data that is processed using the array processors and back projected to produce the digital image. This image is stored and/or displayed for viewing. Additionally, this image can be processed further using special digital image processing algorithms via the image processor.

The events represented in the flow of data are outlined as follows:

(a) The X-ray tube and detectors collect X-ray transmission (data) from the patient.
(b) Detectors measure not only the transmitted photons from the patient but also the intensity of the X-rays from the X-ray tube (referred to as the reference beam).
(c) The transmitted beam and the reference beam are both converted into electrical signals.
(d) These electrical signals are then converted into digital data by the ADCs (analog-to-digital converters).
(e) Data processing involves first preprocessing, after which the data is referred to as raw data.
(f) The raw data are then sent to the array processor for further image processing (convolution – not within the scope of this chapter), and the result is convolved data (the image is not available as yet).
(g) The convolved data is then subjected to another processing operation called back projection, performed by the back projector to produce reconstructed data. This data is the image.
(h) The reconstructed image can be displayed for viewing on a computer screen and archived locally. This image can also be sent to a PACS.
(i) Further processing (digital image processing) can be done on the image via the image processor.
(j) The control terminal is usually an operator's control terminal for complete system control.

5.3.3 CT Windowing: A Postprocessing Operation

The idea of manipulation of the displayed image is referred to image postprocessing. Earlier, we mentioned that the CT image is a matrix of CT numbers and that the range of the CT numbers is referred to as the *window width* (*WW*) and the center of the range is called the *window level* (WL) as is illustrated in Fig. 5.7. We also stated that by manipulating the WW and the WL, we could change the contrast and the brightness of the image, respectively. This operation is a digital image processing operation called windowing.

Windowing in CT uses the CT numbers to manipulate the image gray scale. By manipulating the CT numbers of the various tissues, the image can be changed to

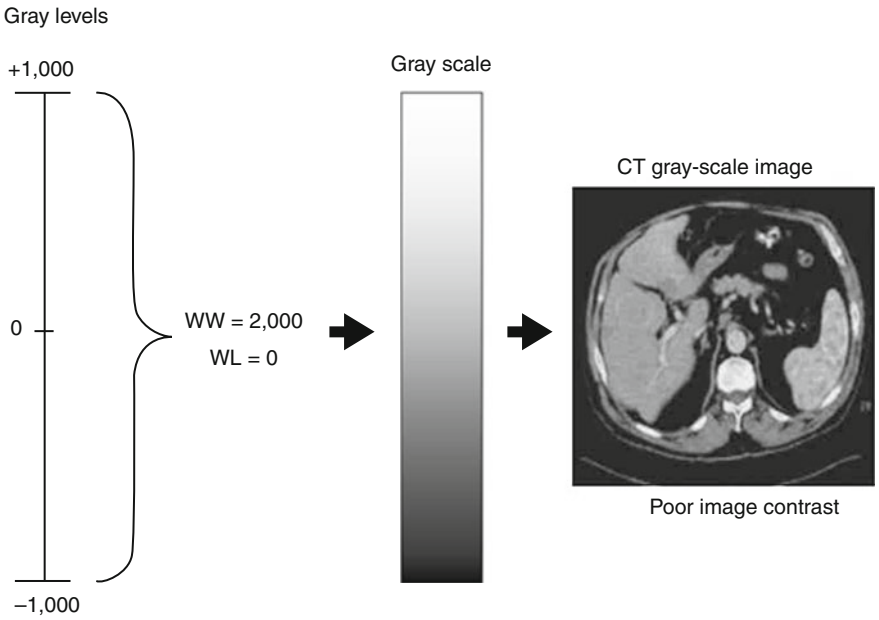


Fig. 5.7 The representation of the concepts of window width (*WW*) and window level (*WL*), two characteristics of the digital image postprocessing operation called windowing. *CT* computed tomography (Reproduced, by permission, from Seeram [2]. Copyright © 2009 by Saunders, an imprint of Elsevier Inc.)

show soft tissues, such as the brain, and dense structures such as bone. Figure 5.8 provides a graphic illustration of the effect of a wide WW (2000) window on the grayscale appearance of the CT image, while Fig. 5.8 shows the effect on image contrast when the WW is narrowed to 1,000. In Fig. 5.8, the WL is moved up the scale to a value of 500, and as can be seen, the picture becomes darker. The effect of the WL on picture brightness is clearly illustrated in Fig. 5.9.

By manipulating the WW and WL controls, the operator can change the image quality in terms of its contrast and brightness. WW controls image contrast; the WL controls image brightness. As the WW increases, the contrast decreases. As the WL increases, the brightness of the image decreases since more of the lower CT numbers are displayed. Finally, these windows can be moved along the scale to optimize views of particular anatomical structures in the image.

5.3.4 CT Image Quality: An Overview

There are several characteristics of the CT image that are of significance to the operator and observer for accurate diagnostic interpretation. These characteristics determine the quality of the CT image.

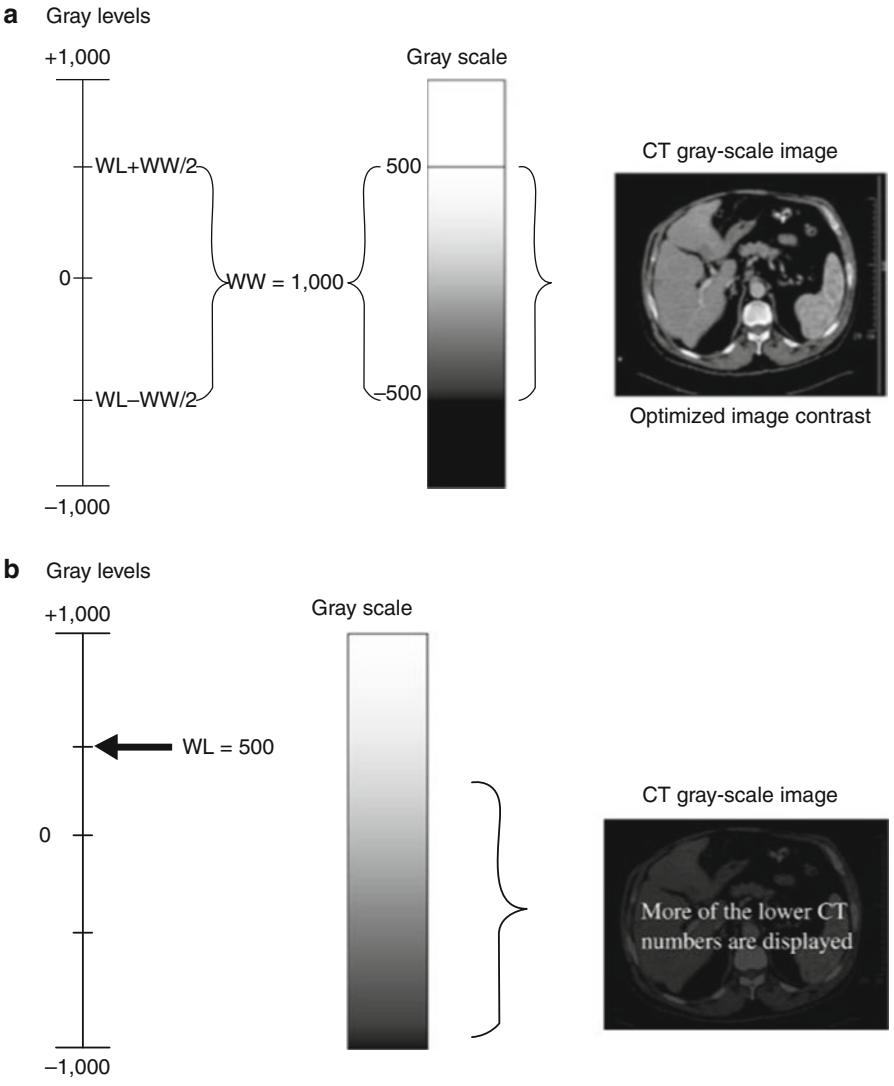


Fig. 5.8 A graphic illustration of the visual effect of window width (*WW*) and window level (*WL*) on image contrast and brightness, respectively. A narrow WW will improve the image contrast (**a**), whereas a higher WL will decrease the image brightness (**b**). *CT* computed tomography (Reproduced, by permission, from Seeram [2]. Copyright © 2009 by Saunders, an imprint of Elsevier Inc.)

The five principal characteristics of CT image quality are spatial resolution, contrast resolution, noise, linearity, and uniformity. In this chapter, only the first three are relevant and will be described briefly.

Spatial resolution is the ability of the CT scanner to faithfully reproduce the details in an object. For CT, the modulation transfer function (MTF) is used to

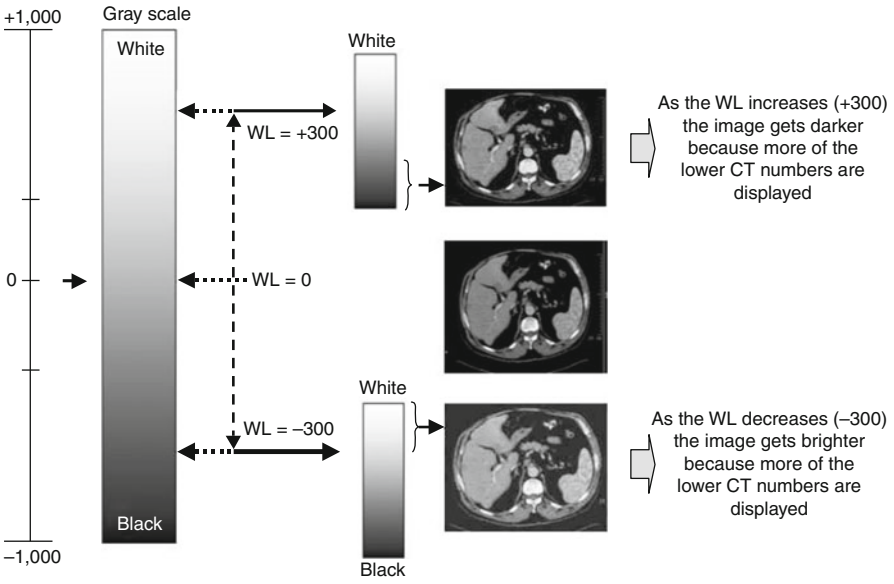


Fig. 5.9 A more dramatic graphical representation of the visual effect of window level (*WL*) on image brightness (Reproduced, by permission, from Seeram [2]. Copyright © 2009 by Saunders, an imprint of Elsevier Inc.)

measure this detail. The MTF is a mathematical function. For our purposes, the MTF is a graph showing the image fidelity as a function of spatial frequency, expressed in line pairs per cm (lp/cm). An MTF of 1 would be a perfect reproduction of fine details in the object; however, many CT scanners are assessed by the spatial frequency at an MTF of 0.1, which is sometimes called the limiting resolution. The higher the line pairs/cm at an MTF of 0.1, the better the spatial resolution of the scanner.

A simplified approach to understanding spatial resolution of a digital image such as the CT image is to use the format of the image. Recall that the CT image is made up of a matrix of numbers. Each of the numbers represents tissue characteristics based on attenuation data and is displayed as a pixel on the image. The image is therefore made up of pixels. The size of the pixel determines the spatial resolution or sharpness of the image, among other factors. The smaller the pixel, the sharper is the image.

Before the CT examination, the size of the matrix is chosen depending on the anatomy under investigation. The operator must select the field-of-view (FOV) or reconstruction circle, which is the circular region from which the X-ray transmission measurements are recorded during scanning. The pixel size can be computed from the FOV and the matrix size using the following relationship:

$$\text{Pixel size} = \text{FOV} / \text{matrix size}$$

For example, if the reconstruction circle (FOV) is 25 cm and the matrix size is 512 × 512, the pixel size can be determined as follows:

$$\begin{aligned}\text{Pixel size} &= 25 \times 10\ \text{mm}/512\\ &= 250\ \text{mm}/512\\ &= 0.49\ \text{mm}\\ &= 0.5\ \text{mm}.\end{aligned}$$

Finally, another important point regarding the format of the CT image is that each pixel in the CT image can have a range of gray shades, sometimes referred to by some authors as the *dynamic range*. For example, an image can have 512 (2^9), 1,024 (2^{10}), or 2,048 (2^{11}) different grayscale values. Since these numbers are represented as bits, the CT image is often characterized by the number of bits per pixel. This is referred to as the *bit depth*. The numeric value of the pixel represents the brightness of the image at that pixel position. A CT image with 12 bits per pixel would represent numbers ranging from 1,000 to 3,095 for a total of 4,096 (2^{12}) different shades of gray.

The *contrast resolution* of CT is significantly better than film-screen radiography. This is one of the advantages of CT compared to radiography. Recall that contrast resolution is the ability of the imaging system (detector) to show small differences in tissue contrast. In CT, because special electronic detectors are used and because the CT image is calculated and converted to a grayscale image, tissue contrast differences of 0.1 % can be detected. Additionally, because of the narrow beam collimation used in CT, scattered radiation is reduced, thus improving contrast.

Noise in CT is a variation in CT numbers from pixel to pixel in the CT image matrix. For example, if a water phantom is imaged, all pixels should have a 0 value (CT/Hounsfield Unit number for water is 0). The variation of these numbers is noise. Noise can be caused by the fluctuation of photons at the detector. This is the source of quantum noise. Noise is also influenced by several factors such as the kV, beam filtration, FOV, pixel size, slice thickness, algorithm, detector efficiency, and patient dose.

5.3.5 *Multi-slice CT: Fundamental Physical Principles*

Current state-of-the-art CT scanners are multi-slice (MSCT)/multi-detector. This technology has become commonplace, and therefore, a discussion of the general principles is in order.

While the earlier CT scanners are based on scanning one slice at a time (single-slice CT), other approaches are needed in order to increase the volume coverage speed. These scanners are called volume scanners since a volume of tissue is scanned in a single breath-hold for one revolution of the X-ray tube and detectors. To achieve volume scanning, a method had to be developed to allow the X-ray tube and detectors to rotate continuously while the patient moves through the scanner gantry to cover the volume of tissue required. This challenge was met by the use of slip-ring technology as well as new detector technology.

Presently two types of volume CT systems are available for use in clinical settings. These include single-slice volume CT scanners (SSCT) and multi-slice volume CT scanners (MSCT), although SSCT scanners are becoming obsolete.

5.3.6 Conventional CT Scanning: Shortcomings

The problems imposed by CCTS are several and have provided the motivation to develop faster CT scanners. These shortcomings have been described by Seeram [2] and include longer examination time due to the interscan delay (ISD), cable wraparound, slice-to-slice misregistration, and inaccurate generation of 3D images. The ISD refers to the time delay that is necessary to reposition the patient for the next slice due to the fact that as the X-ray tube rotates around the patient, the high-voltage cable follows it and wraps around the gantry. This cable wraparound prevents the X-ray tube from rotating further, and so the process stops. The patient is then positioned for the next slice, and the scanning continues with the X-ray tube rotating in the opposite direction to unwind the cable wraparound. This method of scanning has also been referred to as a stop-and-go method. Slice-to-slice misregistration refers to problems of missing portions of anatomy due to the repositioning for the next slice or due to the inconsistency of respiration on the part of the patient. The slice-by-slice data acquisition also produces poor 3D reconstructed images which appear blocky appearance of the 3D image, and these 3D images are said to demonstrate the stair-step artifact. These limitations are overcome by volume CT scanning using spiral/helical beam geometry.

5.3.7 MSCT Scanning: Basic Principles and Technology

Volume scanning evolved from SSCT to MSCT. Therefore, it is essential that we review the basics of SSCT before we describe the essential elements of MSCT. The requirements for CT scanning of a volume of tissue continuously are several and include the use of slip-ring technology, continuous table movement, increased X-ray tube loadability, increased X-ray tube cooling capability, spiral/helical interpolation algorithm, and mass computer memory buffer. While it is not the focus of this chapter to describe the details of each of the above, it is essential to explain briefly slip-ring technology and the spiral/helical interpolation algorithm.

5.3.8 SSCT: Overview

There are two steps in volume scanning that are different than CCTS. These relate to the data acquisition and image reconstruction. First, to scan a volume of tissue, it must be possible to have the X-ray tube and detectors rotate continuously around the

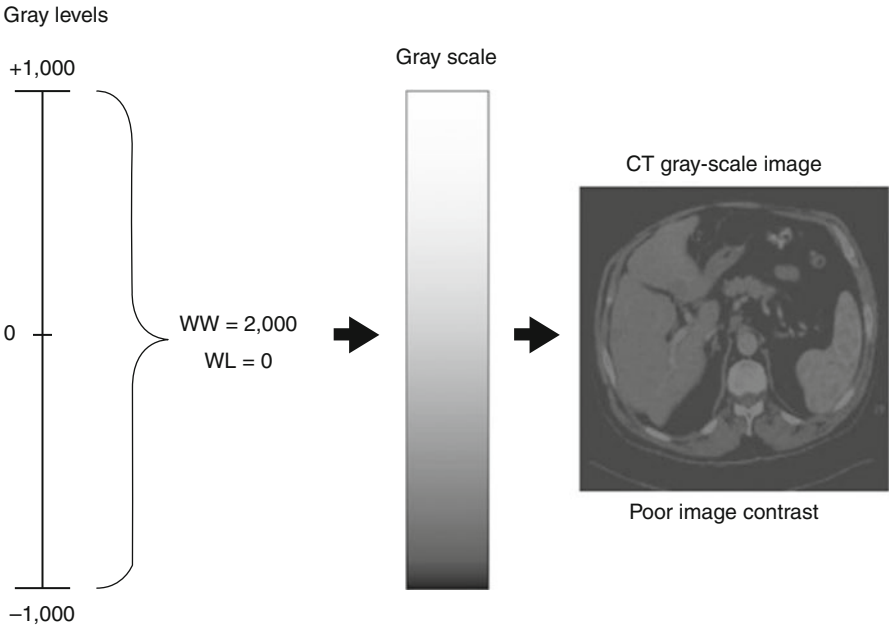


Fig. 5.10 In early spiral/helical volume computed tomography (*CT*) scanners, the use of an interpolation algorithm is an essential first step to compute a planar section, from which all other axial sections are generated (Copyright © 2009 by Saunders, an imprint of Elsevier Inc.)

patient, while the patient moves simultaneously through the gantry. Such a task is accomplished by using slip-ring technology. Slip rings are electromechanical devices consisting of circular electrical conductive rings and brushes that transmit electrical energy across a rotating interface [2]. Slip rings remove the cable wraparound problem of CCTS by providing electrical energy to the X-ray tube that is mounted on the rotating frame of the gantry. The X-ray generator is also mounted on this frame and uses a short piece of high-voltage cable connected to the X-ray tube.

The process for image formation in volume CT is illustrated in Fig. 5.10. Data acquisition, as you will recall, refers to the method of collecting X-ray transmission readings (attenuation data) from the patient. Volume CT is characterized by the rotation of the X-ray tube continuously while the patient moves simultaneously through the gantry. There are several problems resulting from the movement of the patient while the X-ray tube is energized. First, there is no defined slice (as in CCTS), and localization of a specific slice is difficult. Secondly, the geometry of the slice volume is somewhat different than in CCTS. In addition, the effective slice thickness increases because it is affected by the width of the fan-beam of X-rays and the speed of the table. Fourthly, because there is no defined slice, the projection data are inconsistent. This point is important because consistent data are needed to satisfy the filtered back-projection algorithm. Finally, if the inconsistent data are used with this algorithm, streak artifacts similar to motion artifacts are clearly seen on the image.

To overcome these problems, the following image reconstruction steps are used. First, it is necessary to produce a planar section similar to the slice in CCTS. Such a planar section is generated using interpolation algorithms. Interpolation is a mathematical technique to estimate the value of a function from known values on either side of the function. It is not within the scope of this chapter to discuss these interpolation algorithms. For a complete description of these algorithms, the interested reader should refer to Seeram [2].

Secondly, the images are reconstructed using the standard filtered back-projection algorithm. The results are CT images free of motion artifacts.

There are a few equipment-related differences between SSCT scanners and MSCT scanners. In addition, MSCT makes use of a set of parameters that are unique only to volume CT scanning. In SSCT, the detector system is a 1D detector array. This system collects one slice per revolution of the X-ray tube and detectors. In MSCT, the detector system is a 2D detector array consisting of varying rows. An MSCT detector with 4 rows will collect four slices per revolution. One with 16 rows or 64 rows will collect 16 or 64 slices per revolution of the X-ray tube, respectively. Hence, MSCT increases the volume coverage speed by virtue of its detector design.

The set of parameters that are unique to both SSCT and MSCT are pitch, pitch ratio, volume coverage, table speed, and reconstruction increment.

The term *pitch* refers to the distance the table travels per rotation of the X-ray tube and detectors. The *pitch ratio* on the other hand is defined as the ratio of the distance the table travels per 360° rotation to the slice thickness or beam collimation. A pitch ratio of 1:1 is equivalent to scanning using a conventional CT scanner. SSCT uses pitch ratios higher than 1:1 (e.g., 2:1 and 3:1) to increase the volume coverage speed; however, increasing the pitch ratio tends to decrease image quality.

Volume coverage is a ratio of the product of scan time, collimation, and pitch to the gantry rotation time. The *table speed* or table increment is the distance (mm) the table travels per second. Finally, the *reconstruction increment* (also referred to as the reconstruction interval or reconstruction spacing) determines the degree of sectional overlap needed to improve image quality.

One of the major problems with SSCT is its limited volume coverage speed. If the pitch is increased to increase the volume coverage speed, then image quality is degraded. MSCT scanning increases the volume coverage speed without a degradation of image quality.

5.3.9 Multi-slice Detectors

One major problem with single-slice, single-row detectors is its limited volume coverage speed. In 1992, Elscint introduced the first dual-slice (2 detector rows) volume CT scanner to increase the volume coverage speed and thus decrease the time for data collection. CT scanners now use multirow detectors to image multi-slices during a 360° rotation. It is important to realize that other terms such as multi-detector and multichannel have been used to describe the detectors for MSCT scanners.

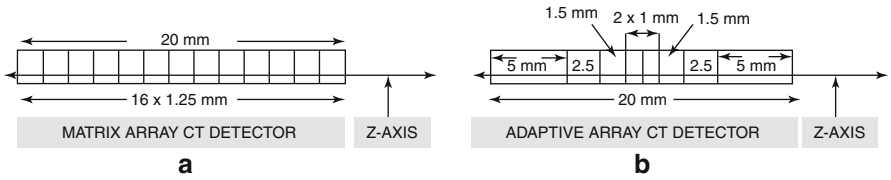


Fig. 5.11 Design elements of three types of detectors for multi-slice computed tomography scanners. See text for further description

The goal of MSCT detectors is to increase the volume coverage speed performance of both single-slice and dual-slice CT scanners. The MSCT detector consists of a 2D detector array with rows of detector elements. A detector with *n* rows will be *n* times faster than its single-row counterpart. These detectors are solid-state detectors that can acquire 4 to 64 to 320 slices per 360°. An MSCT detector consists of multiple separate detector rows.

MSCT detectors fall into two categories (Fig. 5.11), namely, *matrix array detectors* and *adaptive array detectors*. The matrix array detector (Fig. 5.11, A) is sometimes referred to as a fixed array detector, contains channels or cells as they are often referred to, that are equal in all dimensions. Because of this, these detectors are sometimes referred to as being *isotropic* in design, that is, all cells are perfect cubes. The adaptive array detector on the other hand is *anisotropic* in design. This means that the cells are not equal but rather they have different sizes (Fig. 5.11). The overall goal of isotropic imaging is to produce improved spatial resolution in both the longitudinal and transverse planes. During scanning, the number of slices and the thickness of each slice are determined by the detector configuration used. For the sake of simplicity, the detector configuration for a 4-row matrix array detector is illustrated in Fig. 5.12. In this example, each detector channel is 1.25 mm, and 4 cells are activated or grouped together to produce 4 separate images of 1.25 mm thickness per 360° rotation. On the other hand, 8 cells can be configured to produce 4 images of 2.5 mm thickness (1.25 mm + 1.25 mm = 2.5 mm) per 360° rotation and so on. Multirow detectors feature a number of imaging characteristics that are important to the technologist during scanning, and the interested reader should refer to Seeram [2] for further details.

5.3.10 Selectable Scan Parameters

There are several scan parameters in CT that can be selected by the operator, and these include the scan mode, exposure factors (kV, mA, and scan time), gantry rotation time, pitch, scan length, collimation, and slice width. However, only the pitch ratio (often referred to as pitch for short) will be elaborated here to impress upon the reader that there has been much debate about the actual definition of pitch as the technology for CT scanners evolved. In SSCT scanning, pitch is defined as the distance the table travels per rotation of the X-ray tube, T (mm), to the width of the X-ray beam, W (mm). This can be stated specifically as

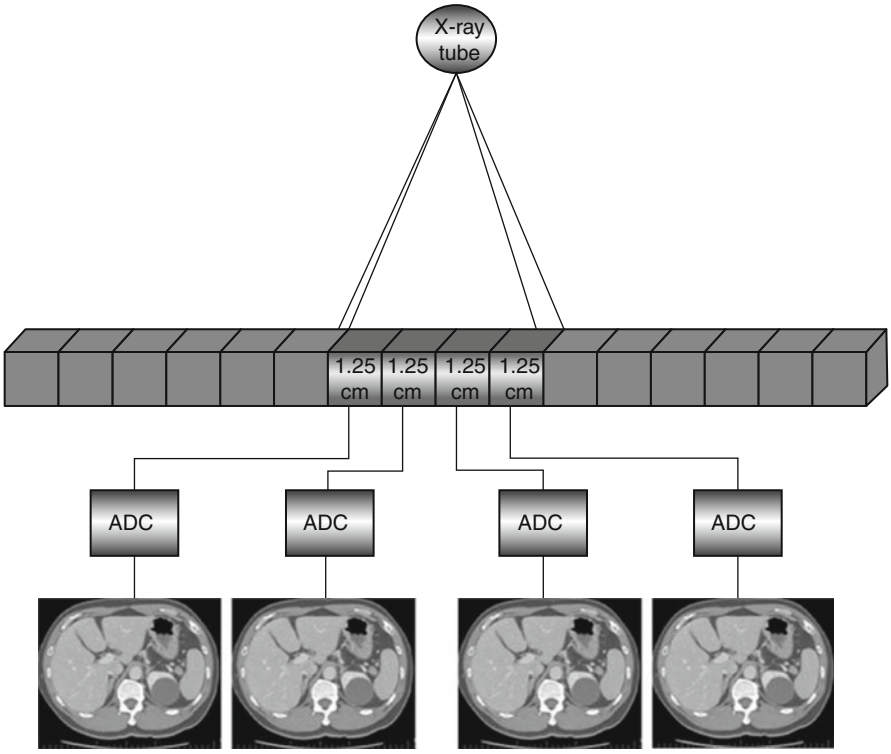


Fig. 5.12 The detector configuration for a four-row matrix array detector. In this example, each detector channel is 1.25 mm, and four cells are activated or grouped together to produce four separate images of 1.25 mm thickness per 360° rotation. On the other hand, eight cells can be configured to produce four images of 2.5 mm thickness (Reproduced, with permission, from Seeram [2]. Copyright © 2009 by Saunders, an imprint of Elsevier Inc.)

$$\text{Pitch}_{\text{SSCT}} = \text{Beam Pitch} = T/W$$

For MSCT scanning that features a 2D detector array system with multiple detectors rows, each detector row has a width of D (mm). Subsequently, the following definition of pitch has emerged for these scanners and stated mathematically as follows:

$$\text{Pitch}_{\text{MSCT}} = \text{Detector Pitch} = T/D$$

While SSCT uses the term "beam pitch," MSCT uses the term "detector pitch." The concept of beam pitch can be applied to MSCT scanning as follows:

$$\text{Beam Pitch}_{\text{MSCT}} = \text{Detector Pitch}/N$$

where N is the number of active detector rows, that is, the X-ray beam is collimated to N detector rows.

5.3.11 The Current Definition of Pitch: The IEC Definition

The current and universal definition of the term pitch has been an activity of the International Electrotechnical Commission (IEC), and this group states that the pitch (P) is equal to the distance the table travels per rotation (d)/total collimation (W). The total collimation on the other hand is equal to the number of slices (M) times the collimated slice thickness (S). Algebraically, the pitch can now be expressed as

$$P = d/W \text{ or } P = d/M \cdot S$$

5.4 Advantages of MSCT

The advantages of MSCT are as follows:

1. Substantial reduction of examination time
2. Coverage of extended anatomic structures
3. Substantial increase in longitudinal resolution by means of reduced section width
4. Isotropic spatial resolution
5. 3D and MPR images with resolution similar to transverse sections

5.5 Isotropic Imaging

An important objective of increasing number of slices (4–320 slices) per rotation of the X-ray tube and detectors around the patient is to achieve isotropic imaging which simply refers to the size of the voxels used in a volume data set. When the voxel is a perfect cube, the data set acquired is said to be isotropic. If, however, all voxel dimensions are not equal, that is, the slice thickness is not equal to the pixel size, the data set acquired is said to be anisotropic.

The purpose of isotropic imaging in CT is to achieve excellent spatial resolution (detail) in all imaging planes, especially in multiplanar reconstruction (MPR) and 3D imaging. Therefore, in MSCT scanning, personnel should have a working knowledge of how voxel size affects not only the spatial resolution but also the radiation dose. Such understanding ensures that personnel work within the ALARA philosophy in order to optimize the image quality and radiation dose.

One significant factor that affects isotropic imaging is the detector configuration, that is, how the detector elements are used together with the effective slice thickness.

MSCT scanners from 16-, 32-, 40-, and 64-channel MSCT scanners can produce isotropic voxels and hence achieve isotropy. MSCT scanners (beyond 4 slices) can also produce voxels that are anisotropic.

5.6 Applications of Multi-slice CT Scanning

Multi-slice volume CT scanning has opened up new avenues in CT imaging, such as improvements in CT fluoroscopy (CTF), computed tomography angiography (CTA), three-dimensional (3D) imaging, and virtual reality (VR) imaging, of which CT endoscopy is a principal example. While it is beyond the scope of this chapter to describe the full details of how each of the techniques works, it is noteworthy to outline their basic principles of just 3D and virtual reality imaging (VRI) since they are becoming commonplace. *3D imaging* provides additional details to allow for qualitative and quantitative evaluation of the large amounts of data collected in CT (as is characteristic of MSCT).

The data acquired from the patient in CT are stored in 3D space, a space defined by the use of a coordinate system to describe an object by measuring distances from a defined point of intersection or zero point. Computer software can be used to aid an observer to view all aspects of 3D space. This is referred to as 3D visualization, and the application of 3D visualization techniques in medicine is referred to as 3D medical imaging.

VRI is one area of extensive research since it is intended to facilitate the perception of 3D anatomy from a set of 2D images. The vast amount of data collected by MSVCT scanners provides the opportunity to develop VRI imaging applications in radiology. *Virtual reality* is a branch of computer science that immerses the users in a computer-generated environment and allows them to interact with 3D scenes. One common method that utilizes virtual reality concepts is virtual endoscopy, popularly referred to as CT endoscopy. CT endoscopy is used to create inner views of tubular structures. More recently *cardiac CT imaging* is now possible with the development and use of scanners designed specifically to image the beating heart.

One of the problems of the previous MSCT scanners using 4–64 slices per revolution however is related to obtaining dynamic 3D images of the beating heart in cardiac imaging. Furthermore, these MSCT scanners create image artifacts due to the beating heart and have limited organ coverage due to the size of the detector (20–40 mm). The latter implies that two or more rotations are needed to cover the entire organ such as the heart or lungs. To address these limitations, two other MSCT scanners are now commercially available and are dedicated to imaging the heart. These scanners are the 320-Slice Dynamic Volume CT Scanner and the Dual-Source CT Scanner. For a detailed description of the elements of these scanners, the interested reader should refer to Seeram [2].

Furthermore, MSCT scanners have found other applications in a number of clinical arenas, including radiation therapy, nuclear medicine, and breast imaging. In

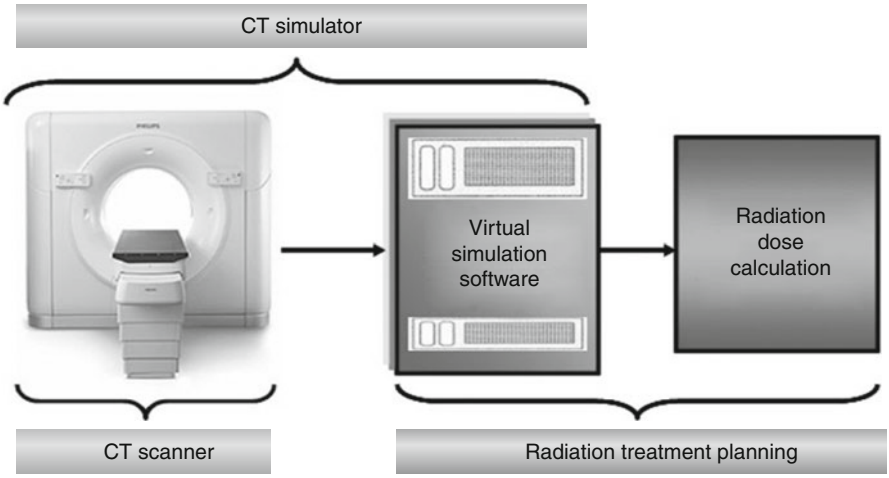


Fig. 5.13 The use of the computed tomography (*CT*) scanner in radiation therapy. As shown, the CT scanner is coupled to virtual simulation software to create the CT simulator to provide data for radiation dose calculations (Reproduced, with permission, from Seeram [2]. Copyright © 2009 by Saunders, an imprint of Elsevier Inc.)

radiation therapy, for example, the CT scanner is now used in the radiation oncology department and plays a role in the radiation treatment planning process, and it is an integral part of the CT simulation process. CT simulation is a technique whereby dose beam arrangements and treatment fields are simulated without any information of the dose. As shown in Fig. 5.13, the CT simulator is coupled to the radiation treatment planning system to provide data for radiation dose calculation.

The CT simulator is a CT scanner characterized by physical devices (hardware) and specialized software. While the hardware is a MSCT scanner featuring a flat tabletop (to ensure that patients are scanned in exactly the same position as they would be when they are on the radiation treatment machine) and other important devices, specialized software (virtual simulation software) is used to define any simulation based on the software-created virtual simulator and a volumetric patient scan. The scan does not necessarily have to be a CT, and other imaging modalities can be used [3]. The CT simulation process includes at least three steps that involve scanning the patient in the CT scanner, planning the treatment and CT simulation, and finally setting up the patient in the treatment machine.

Yet another application of MSCT is image fusion. This is receiving attention in the literature since anatomical images such as those from the CT scanners, MR scanners, and functional images from nuclear medicine (e.g., positron emission tomography (PET) and single-photon emission computed tomography (SPECT)) are now used not only in radiation therapy but in nuclear medicine (SPECT-CT and PET-CT), to enhance the medical management of the radiation therapy patient. The basic steps of image fusion are illustrated in Fig. 5.14.

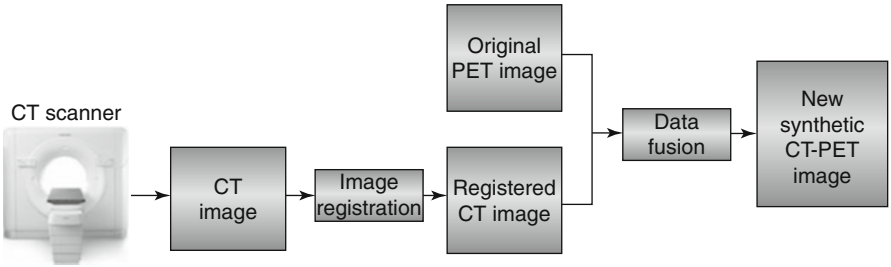


Fig. 5.14 Image fusion uses anatomical images from a computed tomography (*CT*) scanner as well as functional images from nuclear medicine (positron emission tomography [*PET*] and single-photon emission computed tomography) to create a new synthesized image that can be used to provide additional information for enhance diagnosis (Reproduced, with permission, from Seeram [2]. Copyright © 2009 by Saunders, an imprint of Elsevier Inc.)

Recently, CT mammography has sparked interest once again, and efforts are on the way to develop breast CT scanners. The basic design framework for an FD-CT scanner for breast imaging what is referred to as a pendant geometry cone beam CT imaging system, where the patient is placed in a prone position on the tabletop and the breast to be scanned is inserted through the hole to hang during the imaging process using a cone beam wide enough to cover the detector. The pendant geometry prevents exposure of the chest cavity. The X-ray tube and detector are positioned close to the underside of the tabletop and rotate around the hanging breast. X-ray transmission readings are collected, digitized, and subsequently sent to the computer for image reconstruction using special reconstruction algorithms.

5.7 CT and PACS

Interfacing a CT scanner to a picture archiving and communication system (PACS) is an enormous task. There are several issues that need consideration, for example, the CT workstation, worklists, image distribution, Hospital Information Systems (HIS)/Radiology Information Systems (RIS)/PACS Integration, and DICOM specifications. CT workstations will be used primary for diagnosis, and therefore, an immediate concern is to orient radiologists to the nature of this workstation for soft-copy display of images. *Worklists* are used to match cases from the CT scanner to various workstations, and therefore, a PACS must be capable of creating and using worklists effectively. *Image distribution* allows for the system to send images to remote locations such as the radiologist's home where he/she can download them for interpretation.

Another important aspect of CT and PACS is the integration of a CT scanner not only with the PACS but also with the HIS/RIS. Therefore, connectivity is important, and in this regard, the communication standard for medical images (DICOM – Digital Imaging and Communications in Medicine) and more specifically DICOM conformance is mandatory.

5.8 Quality Control in CT

Quality control (QC) is an important part of any department's quality assurance program (QA) and incorporates equipment tests carried out to monitor performance. QC in this scenario is a systematic process to regularly test CT imaging equipment and compare the performance to a known and accepted standard ensuring that the desired quality criteria are met.

5.9 Categories of QC Test

There are two main categories or levels of QC tests carried out on a CT system, and these are acceptance testing and constancy testing. It is necessary to perform acceptance tests following installation of a CT system or after significant modifications have taken place such as software or hardware updates. Constancy tests are carried out on a more frequent basis (daily, weekly, monthly, biannually, or annually). The daily, weekly, and monthly tests are often carried out by a member of staff working in the CT department and designated to perform QC. Further QC tests will be performed by engineers or physicists to coincide with the servicing of the CT system.

5.9.1 Acceptance Tests

The purpose of acceptance testing is twofold in that it validates that the equipment is suitable for clinical use and meets the standards specified by the manufacturer and it also provides reference values against which future routine QC test results can be compared.

Radiation testing is a vital part of acceptance testing and will ensure that the room is adequately shielded and that there is no radiation leakage. Shielding is necessary to ensure protection of staff working in or near to the CT department and also patients and other members of the public who could be closely situated in other areas of the department. Estimation of the shielding required should be made during planning of the installation of a CT system, and its integrity is tested following installation [4].

Mechanical testing will be performed to ensure that there is no possible collision of patient couch and that all the emergency stop devices are working adequately. It is also necessary to consider the correct mechanical registration within the system. This would include testing alignment of the laser lights, accuracy of the couch incrementation and travel, and also precision of gantry tilt. Electrical tests will also be performed at acceptance stage.

Another important assessment is that of the peak tube voltage (kVp) output of the X-ray tube. CT typically involves the use of a high kVp of around 120 kVp. The use of a high kVp provides the photons in the X-ray beam with sufficient energy to

penetrate the range of anatomical structures encountered within a patient. Only photons which reach the detectors are able to contribute to the image produced, so any deviation in kVp is likely to affect both image quality and radiation dose to the patient. If we consider what would happen if X-rays were produced using a low kVp, the photons would have a lower energy and would be less penetrating than when a higher kVp is utilized. Low-energy X-ray photons are more likely to undergo attenuation within the patient and be absorbed by superficial tissues and, as such, are more likely to contribute to increased radiation dose to the skin.

5.9.2 Image Quality

Image quality in CT is a difficult concept to define as it tends to be subjective to the person viewing the image. Subsequently, this subjectivity of the observer and their perception is linked to diagnostic accuracy. It is possible to acquire objective measurements for image quality as is often the case with QC tests. It has to be said that direct correlation of these measurements with those of the observer is uncertain. This implies that minor changes in image quality might not be detectable to the human eye. For example, increase in tube current (mA) would increase contrast resolution in the resultant image and decrease noise. However, above a certain level, this would not be detected by the observer and so add no value to the diagnostic quality of the image with regard to conspicuity but would cause an unnecessary increase in radiation burden to the patient.

Accuracy of CT number can be derived by measurement of a substance of known CT value. In CT, this is typically water which has a known Hounsfield Unit (HU) value of 0 and air which has a HU value of −1,000. These tests are performed as part of acceptance testing and also as part of the routine constancy QC tests which are performed regularly within the department. The aim is to ensure CT number accuracy and to ensure uniformity by taking measurements from a phantom with known homogeneity. The accepted tolerance range of the mean value is 0 ± 4 HU and for the uniformity is ± 0.2 HU [5].

Noise and uniformity tests using a water phantom are discussed in more depth below (see Figs. 5.15 and 5.16).

5.9.3 Linearity

Linearity of a system can be described as the property which is characterized by the output being directly proportional to the input. In CT, this is the linear relationship between the calculated CT number (output value) and the linear attenuation coefficient of element within a test object (input value). Theoretically, a change in attenuation coefficient should result in a change in CT number, but on the whole, this concept does not always hold true for CT values.

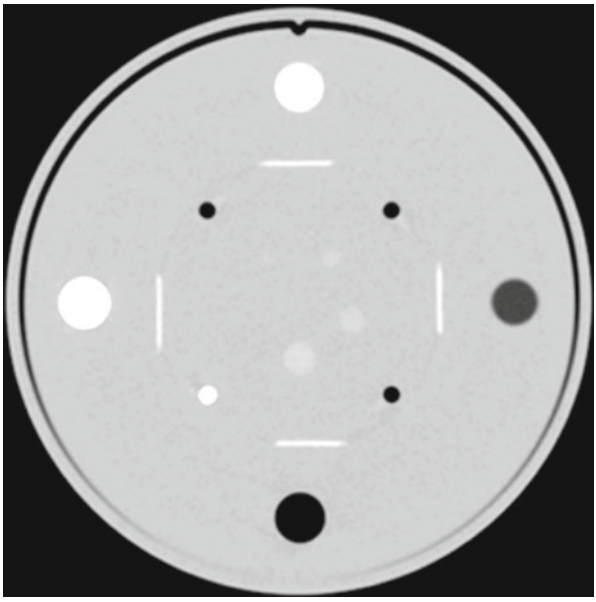


Fig. 5.15 A suitable test phantom for the determination of linearity. The phantom is water filled and contains inserts of various densities, typically ranging from −1,000 to + 1,000 HU

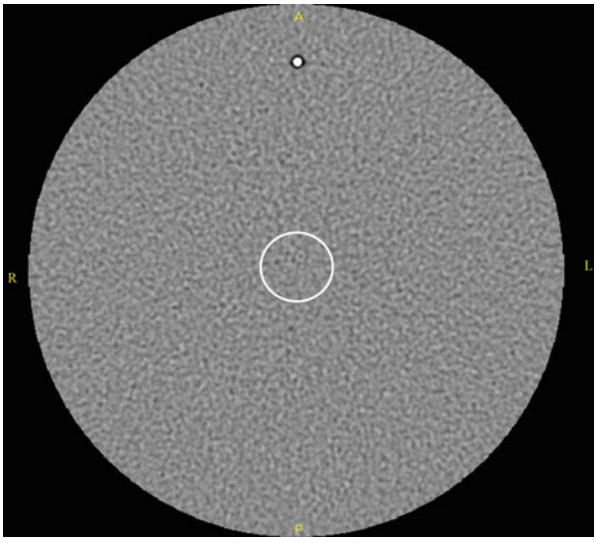


Fig. 5.16 The water phantom being used to assess the noise level of the image by assessing the standard deviation within the central ROI

CT linearity is assessed using a phantom containing a number inserts of different densities which cover a wide range of CT numbers ranging from approximately −1,000 to +1,000 HU [2]. This would represent the range of density values encountered in the human body. An example of a suitable phantom is shown in Fig. 5.17. Linearity should not deviate from the known CT numbers of the elements by more than<± 5 HU. If measures fall outside the acceptable range, then remedial action should be taken [5].

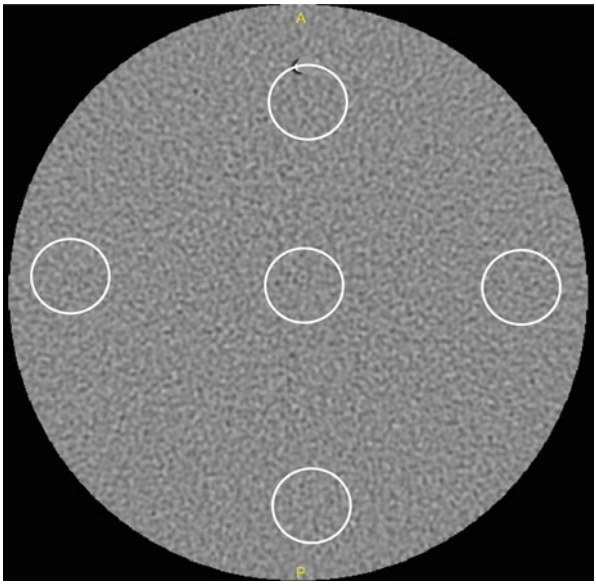


Fig. 5.17 A demonstration of how the water phantom can be used to determine uniformity

Measurements to assess low-contrast resolution (low-contrast detectability) and high-contrast spatial resolution are performed as part of acceptance testing. As they are also repeated during constancy testing, they are discussed in more depth in the following section.

5.9.4 Patient Dose

CT Dose Index (CTDI) should be measured in air under various conditions to include all clinical kV settings, slice widths, beam filters, and a range of mA settings. These measurements can only be acquired at a single-slice setting, and helical CTDI cannot be measured directly. A correction factor is used to convert CTDI data from axial to helical. This is determined by scanning the full length of an ionization chamber incrementally to obtain contiguous axial slices and then in helical mode with a pitch of 1. The ratio of the two doses is equal to the correction factor.

5.9.5 Constancy Tests

5.9.5.1 Daily

CT X-ray tubes have been designed to dissipate heat very efficiently to allow the higher tube loading required for modern multi-slice scanners. When the scanner

is idle for any length of time, the X-ray tube cools rapidly and will need to be warmed up prior to further scanning. Maintaining the X-ray tube at a suitable temperature helps to prolong its life. Warm-up procedures are performed when the scanner is first switched on, and this can be repeated as necessary throughout the day when the tube has been allowed to cool below the manufacturer's recommended level.

During the initial warm-up procedure, most scanners are also programmed to perform an air calibration scan. To ensure that the system is calibrated to air, it is important to make sure that the patient couch is outside the gantry and that the gantry aperture remains free from any obstruction.

A very important part of daily QC which is very rarely mentioned in QC programs but is probably carried out subconsciously by most CT operators is that of observation. A quick visual check of the equipment could reveal certain mechanical damage or faults which while they might not be immediately detrimental to the CT service could warrant follow-up and prevent remedial action being taken in the future.

5.9.5.2 Weekly

When CT is used as part of a hybrid imaging system such as SPECT-CT, it is essential that the two systems produce data that cannot only be used independently but can also be co-registered. A significant feature of constancy QC is the assessment of CT to SPECT registration. If registration is not accurate between the two systems, then misregistration artifacts will be evident on the resultant images. It is probably sufficient to perform registration tests no more frequently than once a week.

Further weekly tests on the CT system are performed using a water phantom to collect data. This test will be performed to assess noise and uniformity. Noise (often referred to as quantum noise) is the degradation of an image resulting from the mottled effect produced in the image as a result of a decrease in signal-to-noise (SNR) ratio. SNR is dependent on the number of photons reaching the detectors and contributing to the image; the higher the number of photons hitting the detectors, the greater the signal and the lower the noise and vice versa.

Noise is measured by the standard deviation of CT number within a region of interest (ROI) which is known to be homogenous, in this case water. The number of photons reaching the detectors is determined by the tube current (mA). An increase in mA increases the number of photons reaching the detectors and so reduces the noise level in the image. However, radiation dose to the patient is increased with increasing mA.

To assess the noise, an axial image of the water phantom is produced and a ROI placed in the center of the image. The noise is given by the standard deviation of the CT number within the ROI.

Uniformity can be assessed using the same axial image. Four further ROIs are placed peripherally on the image in the A, P, R, and L regions (see Fig. 5.4).

The standard deviation of the mean CT number of each of the four peripheral ROIs against the central ROI is calculated to give the uniformity value. As previously mentioned, the accepted tolerance range for uniformity is ± 2 HU [2].

5.9.5.3 Monthly

It is not necessary to perform image quality tests more frequently than every month. These tests will also be performed on a regular basis when the CT scanner undergoes routine service. The regularity of service dates will be dependent upon the contract held between the hospital and the manufacturer. Tests performed as part of image quality should include low-contrast resolution also known as low-contrast detectability and high-contrast spatial resolution.

Low-Contrast Resolution

Contrast resolution is the ability to differentiate between two adjacent structures or tissues of similar density. Contrast resolution is greatly degraded by noise within an image which makes measurement of signal-to-noise ratio (SNR) of an image the best descriptor of contrast resolution.

Spatial Resolution

Spatial resolution is the ability to distinguish two small closely situated objects. The spatial resolution of a system is measured by the delineation of objects as they become closer together. The closer together structures in an image can be demonstrated, the better the spatial resolution.

Although the modulation transfer function (MTF) can be used to describe resolution of a CT system, more commonly test phantoms are used, and contrast and spatial resolution are often a subjective measurement.

5.9.5.4 Annual

Certain constancy tests will be carried out less frequently on an annual basis. Some of the tests performed during acceptance testing will be repeated at this time. Annual tests should be carried out by either engineers or physicists and usually coincide with routine service of the CT system. The tests carried out will include CT number accuracy and uniformity which will have been performed as part of constancy tests throughout the year, CTDI and dose profile, and also high-contrast spatial resolution and low-contrast resolution tests which will also have been performed within the 1-year period. Mechanical registration tests to assess accuracy of the laser light alignment, gantry tilt, and table travel and incrementation and accuracy of slice width will also be performed.

In summary, it is important to determine that a CT system is operating within the required tolerance levels specified by the manufacturer. This involves a variety of tests being performed, some of which have been described above, at varying intervals which can be compared to a standard to ensure that consistency of the

operating system is met over time and that the results of the tests are clearly documented.

5.10 Summary

This chapter has included examples to give an insight into some of the procedures and practices that might occur within a CT or SPECT-CT department but is by no means definitive. Local practice and recommendations by the manufacturer, engineers, and physicists will mean that how and when QC tests are performed might vary between different departments and depending on clinical practices undertaken.

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Part II
Selection and Design

Chapter 6
Selecting an Appropriate SPECT/CT Scanner

Julian MacDonald, Marc Griffiths, and David Wyn Jones

6.1 Introduction

For nuclear medicine departments, the purchase of a gamma camera with CT capability for the first time will require additional consideration and planning. As well as specifying system requirements, there are issues to consider including service continuation during installation, production of a coherent business case, raising capital funds, adhering to organizational procurement policies, and finally commissioning the new system.

Due to professional backgrounds, many nuclear medicine personnel may not have the necessary knowledge or experience to purchase a gamma camera with CT functionality. In addition to the complexities of gamma camera function and design, the CT component adds another dimension for evaluation. CT scanner components have certain design attributes which can affect image data quality and radiation dose to the patient. Also, consideration needs to be given to the use of the system, for instance, whether the hybrid system needs to be capable of producing high-resolution CT images and images with high temporal resolution (as in coronary angiography) or whether it needs to be capable of stand-alone CT imaging as a backup scanner.

It is useful at the initial stage to contact other imaging departments who have experience of using SPECT/CT. This will assist in the formulation of ideas for its use; these thoughts can then be built into the *case of need* and *business case* documents.

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6.2 System Procurement

Initial thoughts and identifying the need, regarding the replacement of an existing system or indeed setting up a new service, should include informal discussions with local imaging and medical staff as to the likely benefit of purchasing a SPECT unit with CT capability. It is advantageous to contact other departments with experience of using SPECT/CT systems to discuss how their service has developed after the introduction of SPECT/CT as this would help direct your thinking to areas which you might not have otherwise considered. It is good practice to raise interest in the new technology and potential capability of a SPECT/CT system by informally inviting manufacturers to give presentations of their systems' capabilities. Consequently, based upon a range of knowledge and experience, the staff responsible for the procurement process can start to begin to formulate informed ideas of what is available and whether the options would meet local operational and clinical demands (Fig. 6.1).

Managers evaluating the cost/benefit of SPECT/CT systems are likely to require evidence of its advantages; they would also require financial information too. When considering this evaluation, the following points should be considered:

- Clinical benefit, for example, increasing the diagnostic value, compared with SPECT alone, in terms of sensitivity and anatomical location of disease. This may have associated cost savings, which should be explained. For example, the surgical excision of parathyroid adenomata is a good example where (a) the precise anatomical location demonstrated by CT fusion can reduce the operation time and (b) the increased sensitivity to aberrant glands reduce the need to reoperate in the case of "missed" tumors and failed treatment. Also, the ability to correct images for SPECT attenuation losses can improve the diagnostic outcomes – and minimize false-positive and false-negative results – which often means further imaging and expense using alternative techniques.
- Increased productivity/innovation – here, redundant SPECT/CT system time can be used for other purposes outside of normal working hours or as normal nuclear medicine workload allows, for instance:
  - CT colonography
  - CT biopsy
  - Conventional CT imaging
  - Trauma imaging
  - Coronary angiography
  - CT coronary calcium scoring
  - Research
  - Combined cancer staging where nuclear medicine investigations can be combined with CT appointments and undertaken by the nuclear medicine staff on the hybrid system

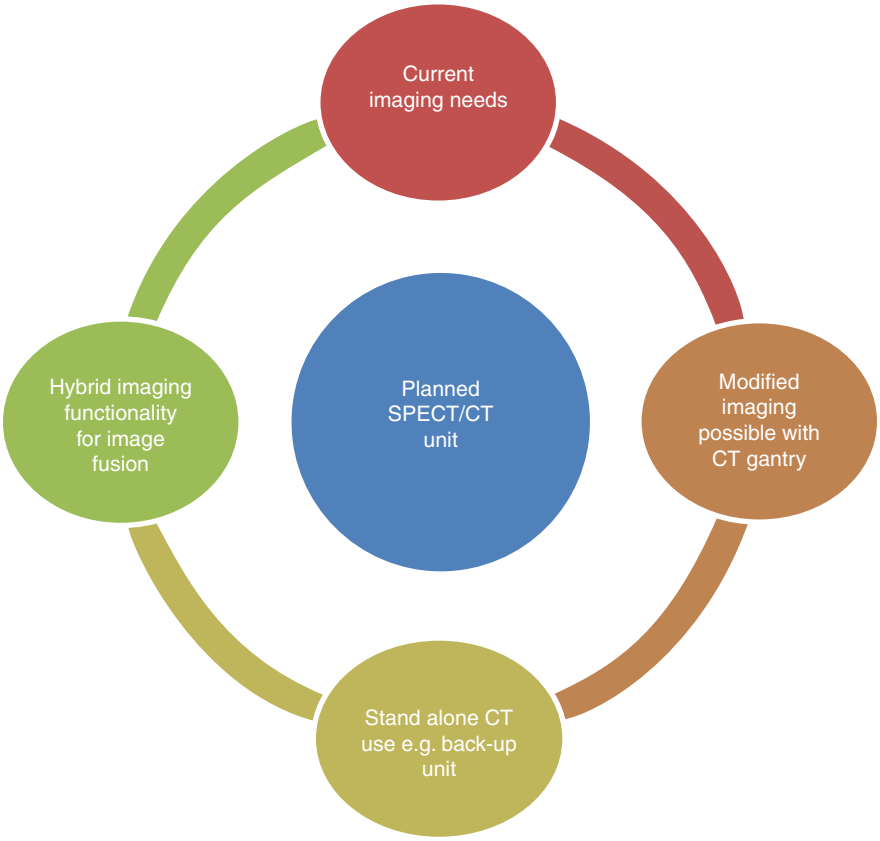

Fig. 6.1 Illustration of some considerations which can assist in specifying a SPECT/CT system to suit local needs in terms of patient flow, type of examinations envisaged, "future-proofing," and formulating innovative ideas for imaging and treatment pathways

- Imaging service resilience – where smaller imaging departments can:
  - Plan to utilize the CT component during routine maintenance of the stand-alone CT scanner used for conventional radiology.
  - In exceptional cases, during unplanned downtime on the radiology CT unit, provide a "backup" service for emergency CT cases.

The above possibilities can be seen as controversial in nuclear imaging terms, where there might be a perceived threat that the CT imaging might overwhelm the existing nuclear service. However, with effective leadership and suitable guidance for service delivery, this potential disadvantage can be turned into a strong advantage in terms of advancing nuclear imaging possibilities with image fusion and

attenuation correction. Of course, the issue of service "creep" toward using CT as stand-alone, if allowed to, will more than likely apply to smaller imaging departments.

Those nuclear imaging departments remote from radiology departments, or indeed where they are separate organizational entities, might benefit from collaborating on some of the above issues in order to procure the best imaging solution for the organization as a whole, bearing in mind the large financial outlay of a HP-CT unit and the possibility for redundant scanner time.

6.3 Formal Procurement

The process of deciding to proceed to plan the purchase of a large and expensive piece of medical equipment, especially one which emits ionizing radiation, has many stages (Fig. 6.2) and usually involves complex local issues together with national/international rules and regulations regarding the procurement process. This is especially true for the publically funded health-care services, such as the British National Health Service (NHS). Most organizations and countries have strict procedures and governmental regulations in the case of large capital investment involving public organizations [1–3]. Consideration also needs to be given to the disposal of old imaging equipment, as this in many countries is regulated too.

At an early stage it is advisable to form a project team involving key individuals likely to have an influence on the procurement, installation, and commissioning of the new system or imaging suite. The project team is likely to consist of the following: a manger, imaging professionals, relevant clinicians, a medical physics expert, a finance representative, and a contracting manager. This team would then collaborate with other departments within the organization and with the "bidding" manufacturers and suppliers – as shown in Fig. 6.3.

In the UK, NHS organizations follow the guidance of the Department of Health (DOH) in terms of the proper conduct for procuring equipment. The following guide represents which might be considered as a logical approach to procurement, based on this advice [4], and it might be generally applicable within other countries. It is advisable however to consult your own policies before proceeding.

Devising a robust business case ensures that there is organizational commitment to deliver the planned change and that there is financial backing to progress with the procurement. The business case also provides a framework for progressing with the project, and it is finally used to monitor the project outcomes to determine whether they have been achieved as set against the business case [4]. The business case document should at the very least address the following [4]:

(a) Strategic fit – description of the business need for purchasing a SPECT/CT system and how it will contribute to the business (imaging) strategy of the organization
(b) Objectives – why it is needed now, key benefits of buying the new imaging system, and the critical success factors and how they will be monitored against the business case objectives

Fig. 6.2 Suggested path towards procuring a SPECT/CT system

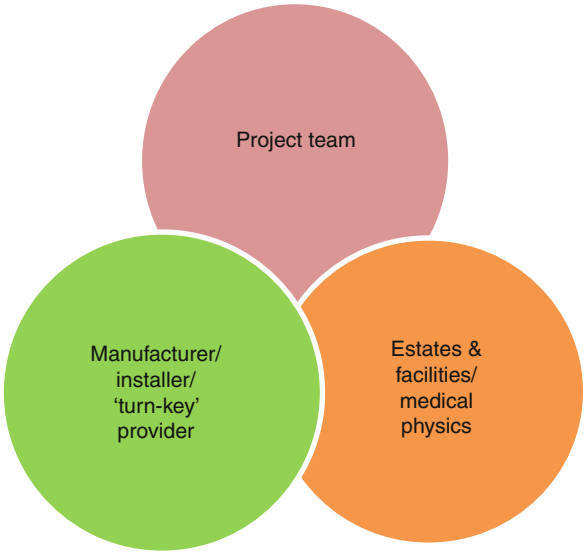

Fig. 6.3 Illustration of collaboration with other departments within the organization and with the "bidding" manufacturers and suppliers

(c) Options appraisal – high-level cost/benefit analysis appraising at least three options and identifying a preferred option with justification for this, for example:

(i) Doing nothing
(ii) Purchasing a system without CT capability
(iii) Purchasing a LP-CT system
(iv) Purchasing a HP-CT system

(d) Commercial aspects – this is usually required where there is an external procurement necessary and should include the following considerations:

  (i) Proposed sourcing options, with a rationale
  (ii) Key features of proposed commercial arrangements (e.g., contract terms, contract length, payment mechanisms, and performance incentives)
  (iii) Procurement approach/strategy, with a rationale

(e) Affordability – to include statements of available funding, and rough estimates of the total cost of the project including future running costs, etc.
(f) Achievability – high-level plan for achieving the desired outcome, with key milestones and major dependencies:

  (i) Outline contingency plans, for example, addressing failure to deliver service on time
  (ii) Major risks identified and outline plan for addressing them
  (iii) Provider's plans for the same, as applicable, skills, and experience required

(g) Source information – containing: project management plans, procurement documentation, high-level requirements, and the business strategy

6.4 Exploring the Market

At an informal initial stage, this can be useful in terms of examining the available SPECT/CT systems. This can be done by visiting manufacturer's commercial websites, informally requesting marketing material, such as brochures and demonstration DVDs, visiting commercial exhibitions and having informal discussions with commercial representatives who can explain system feature. From this a system specification/requirement document can be drawn up, and the formal tendering document can be compiled ready for advertising in the required journals. A good specification should [5]:

- State the requirement completely in a clearly concise and unambiguous manner.
- Primarily focus on outputs (focusing on business deliverables rather than detailed service/system requirements).
- Contain sufficient detail for potential suppliers to determine, and cost, the goods or services they will offer.
- State the criteria for acceptance.
- Provide equal opportunity for all potential suppliers to offer a product or service which satisfies the needs of the organization.

A good specification should not [5]:

- Over-specify requirements.
- Contain features that directly or indirectly discriminate in favor of, or against, any supplier.

The European Union requires all tenders above a specified value to be advertised in the Official Journal of the European Union (OJEU), and other countries are likely to have similar systems for raising interest with the system vendors [1]. Once manufacturers express an interest from the tendering advertisement, expressions of interest are usually forwarded to the project team, even if this is indirectly via the organization's procurement department.

Tender response(s) evaluation can then take place when all the information is to hand from the available suppliers. There should be a robust process in place at this stage to evaluate the responses methodically to compare the responses to the tender requirements. This process will determine the preferred supplier. Consideration should be given to: value for money, project management capability of the supplier, service support (after-sales), engineering, or enabling costs.

The evaluation method should be included in the business case, or at least agreed formally within the project team prior to this stage, so that the goods and services offered meet the required specification set out earlier.

Some tenders can be eliminated as "not meeting the core specification" as set out in the advertised tendering document. These can be filtered out in terms of their capability and capacity to meet the requirement as well as their financial standing [5]. Prequalification questionnaires can be used to achieve this elimination, and will aid any legal process, by fully documenting the procurement decision-making process.

The evaluation of any SPECT/CT system on offer will necessitate:

- The careful and systematic evaluation of supplied technical documentation in support of any bid.
- A technical evaluation of the equipment in a functional clinical setting (site visits) including operation of all system features (although sometimes not all features are available on one site, especially when medical equipment is new to the market, or certain features).
- A formal appraisal of all desirable system features – which can be tabulated in spreadsheet form – later used to systematically compare available shortlisted scanners.
- An equality impact assessment should also be undertaken to meet local policy.

When system features are not available for demonstration or evaluation, it is important that any written contractual agreement details the precise specification and operation required and that the supplied equipment will be constructed and commissioned to fully meet the need, and this clause should be linked to a payment model, for example, some retention of payment, until fully commissioned to specification [5].

The multidisciplinary project team should individually evaluate each system in terms of their own expertise, using an evaluation matrix, and any questions requiring responses from the supplier should be addressed at a single point of contact. Table 6.1 shows an example of an evaluation matrix where scores for features and functionality can be summed. This ensures that there is a good audit trail should problems arise later if a system is installed not quite meeting the expectation of the

Table 6.1 SPECT/diagnostic CT procurement reference site evaluation matrix

Reference site details		
Supplier	System A	System B
Location No. 1	Address A	Address B
Comments, e.g., on type of center and type or work performed on unit	Factory demonstration	Private General Diagnostic center with wide range of investigations using CT/SPECT
Location No. 2		
Comments, e.g., on type of center and type of work performed on unit	Private facility using "x" system only, no CT	NHS Physics run department with full Nuc. Med but limited diagnostic CT usage
Installation and commissioning support		
Quality (adequacy) of application specialist package	4	5
Advice available from commissioning engineer?	0	5
Anticipated installation problems?	5	5
Will CT/gamma camera work independently?	3	4
Overall confidence in system delivering anticipated functionality and reliability?>	4	4
Subtotal:	*16*	*23*
Gamma camera imaging system		
General ease of positioning camera heads (weighted score)	15	20
Gamma camera system overall spec. (tech)	3.5	4
Whole body positioning	4	3.5
SPECT positioning – body	3.5	4
SPECT positioning – brain	3	4
SPECT positioning – cardiac	4	3.5
Ease of "starting" a dynamic acquisition single handed	4	4.5
Closest distance heads can come together	4	4
Ease of positioning transmission phantom, e.g., sentinel node imaging	3	5
Arm rests provided for WB positioning	5	5
Arm rests provided for SPECT	5	5
Head rest supplied for brain SPECT	5	5
Interlocks for touching patients	3.5	4
Crash avoidance system (heads/other equip.)	3.5	4

Table 6.1 (continued)

Reference site details		
Supplier	System A	System B
Termination bell (can it be silenced)	0	0
Visible cue for scan termination (patient)	3	4
Child-friendly features	3	4
Suitability for imaging children overall	17.5	20
Erect positioning in chair, e.g., renograms	3.5	5
Able to image on patient bed?	3	5
Adequate patient weight limit?	5	5
Field of view size for large patients	5	5
Collimator (time to change)	3	4.5
Collimator exchange automation	10	20

Scores 0–5 (0 = no response, 1 = very poor, 2 = poor, 3 = good, 4 = very good, 5 = excellent)

project team [5]. Individual team members should note at this stage the sensitive nature of the information being evaluated, and they should take care to ensure that this stage is confidential. This should guard against releasing commercially sensitive information that might prejudice the whole procurement process.

The project team should also consider the installation and commissioning process as part of the evaluation, including the environment in which the SPECT/CT scanner will be sited. The ability of the scanner to function according to specification might be adversely affected should this part of the process be neglected.

After final consideration, it is sometimes beneficial to have private meetings with each of the companies tendering – in order that final negotiation can take place regarding supplied features within the cost quoted – and to clarify on any ambiguous areas of the procurement process in relation to each system manufacturer's/supplier's goods and services.

The project team will then be in a position to award the contract, with the assistance of procurement professionals within the organization, to the preferred supplier having met most, if not all, of the specification requirements initially set out. Usually this will involve issuing an agreement and cover letter.

6.5 Installation Planning and Building Work

This is often provided as a "turnkey" service by equipment suppliers or can be in-house by hospital estates and facilities department. The essential requirements of a facility to house a SEPCT/CT scanner are discussed in Chap. 7 in detail. However, at this juncture, it is necessary to consider the environment housing the new scanner as being part of the commissioned service, especially if a "turnkey" provider has

installed supporting facilities – as these need careful acceptance checking for compliance with the specification. Local estates and facilities staff can assist in ensuring that the terms of the contract have been met in this respect.

With respect to accepting the new SPECT/CT scanner into use and to ensure that the terms of the procurement are met, the following points need to be evaluated:

- That the equipment supplied is to specification and is the latest model with latest version of software installed.
- That there is no apparent damage.
- That specified accessories are available.
- That any "software packages" are installed and fully functional.
- The system should have local electrical safety checks made to ensure compliance with local legislation.

Thereafter, the following needs to be considered:

- Medical Physics Quality Assurance Acceptance Testing:
  - Ensuring the gamma cameras perform to specification
  - Ensuring the CT scanner and X-ray components perform to specification and within local legal constrains
  - Checks to ensure that the fusion alignment between the CT and gamma camera is to specification
- Radiation Risk assessments should be made – according to local regulation
- Maintaining records of the above
- Ongoing system maintenance contracts
- That equipment documentation is available together with detailed operating instructions and reference material

Staff training on the new SPECT/CT scanner will need to include formal training of local staff by application specialists, and if this training is adequate, then this should be documented as evidence of competency to use a piece of medical equipment. End users need to be aware of the normal operation of the SPECT/CT system in order to use it safely and effectively. Training should cover [1]:

- Any limitations on use
- How to fit accessories and to be aware of how they may increase or limit the use of the device
- How to use any controls appropriately
- The meaning of any displays, indicators, alarms, etc., and how to respond to them
- Requirements for maintenance and decontamination, including cleaning
- Recognize when the device is not working properly and know what to do about it
- Understanding the known pitfalls in the use of the device, including those identified in safety advice from government agencies, manufacturers, and other relevant bodies
- Understanding the importance of reporting device-related adverse incidents to national safety agencies sponsored by government

Finally, the procurement process is likely to be a generalized process internationally, as local legislation is likely to vary in terms of detail. However, the above overview of the process should give a basic understanding of the complexity of the procurement process, some of its pitfalls, and serve to illustrate the need to involve professionals of various professional groups at appropriate stages, to ensure the best outcome.

6.6 Hardware Designs of Modern SPECT/CT Systems

Prior to the introduction of dedicated SPECT/CT systems within clinical practice, a number of techniques were available for the correction of attenuation and/or image registration. In the late 1970s, software-based attenuation correction algorithms began to emerge within medical imaging, as computers became more powerful. Chang's [6] attenuation correction provided a software-based approach to correcting for the effects of attenuation within a relatively uniform field of density (Fig. 6.4).

In the 1980s, increased clinical interest in corrective algorithms and image fusion techniques was again possible due to the provision of powerful dedicated computer systems. However, motion artifacts impacted upon software-based image fusion approaches, which imported reconstructed data from different imaging systems. This was especially the case for clinical areas such as the thorax, abdomen, head, and neck [7]. The use of external fiducial markers has also been utilized within clinical imaging in order to align anatomical and physiological data sets, using relatively basic software [8]. The availability of specialist's three-dimensional elastic transformational/nonlinear warping has emerged within some clinical department

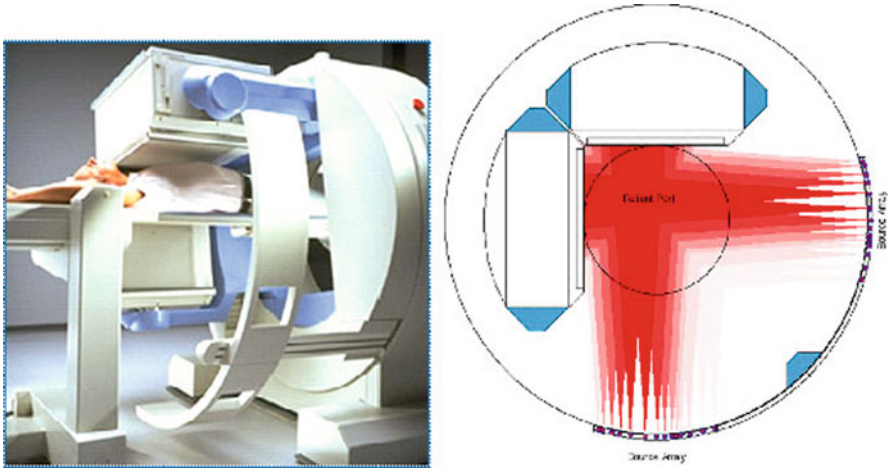

Fig. 6.4 External transmission source array and projected photon flux, which is centered around the myocardium

over the last 10 years, providing a potential solution to the effects of breathing motion artifacts during the registration of two different data sets [9].

Combining data sets from different imaging systems is now considered to be a defined aspect of clinical practice, which is still evolving in many different clinical areas [10]. Published evidence indicates that the level of registration accuracy between data sets (thorax and abdomen) acquired on separate imaging systems is approximately 5–7 mm [9]. Software-based registration of anatomical and physiological data still has an important role in accurately correcting for misregistration issues and for corrective approaches to registration artifacts in the thorax, such as respiratory motion [11].

6.7 External Radionuclide Transmission-Based Correction Systems

The emergence of external radionuclide sources in order to perform attenuation correction became a clinical option in the late 1980s and early 1990s. A range of approaches were adopted, involving the use of moving or stationary radioactive rod sources, configured to provide a transmission map and apply an attenuation coefficient to the specific emission data. Gadolinium-153 was most commonly used within a number of systems [12], which has a physical half-life of 242 days, with two main useful photopeaks of 97 and 103 keV respectfully. Demonstrates a typical external transmission-based correction system, using an array of line profile sources in order to provide attenuation coefficient data, which is then applied to the emission data set.

A number of commercial designs provided the user with an option to perform attenuation correction within myocardial perfusion imaging examinations. Simultaneous emission and transmission imaging is possible using an external-based radionuclide system; however, consideration had to be factored in for the use of thallium-201, where sequential imaging was advised due to potential crossover of gadolinium-153 into the upper energy window of thallium-201. Another limitation of external transmission-based radionuclide correction systems related to the fact that validated use was purely for the purpose of myocardial perfusion imaging. Although transmission maps provided some anatomical information, the spatial resolution was poor (Fig. 6.5) and the photon flux was inferior in comparison to modern CT data.

Issues relating to mechanical misalignment, malfunction of the radionuclide source housing, and ineffective count density values with obese patients have been reported with external transmission source systems [12]. Truncation artifacts are also a potential issue with radionuclide-based transmission sources, especially as size of the patient's body increases. Additional corrective techniques may be employed (i.e., energy window crosstalk correction, scatter correction) to ensure data quality is not affected. It is argued that scatter correction should also be employed in addition to attenuation correction, in order to avoid overestimation of certain areas during myocardial perfusion examinations [13].

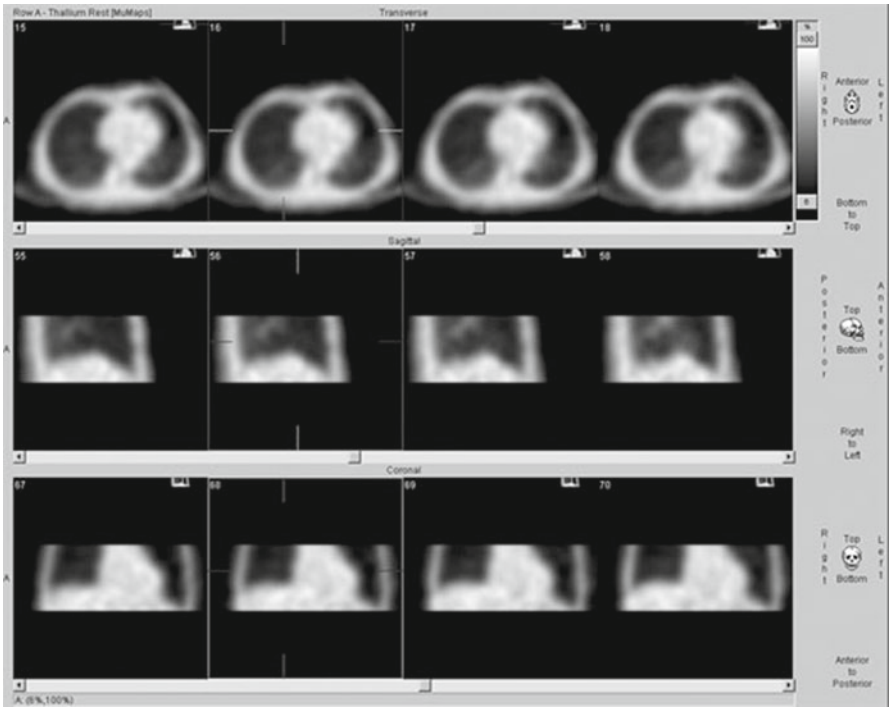

Fig. 6.5 Example reconstructed transmission maps of the lungs, produced from gadolinium-153 sources

Dedicated cardiac gamma cameras are utilized within modern clinical practice, with the option of external radionuclide-based transmission imaging for attenuation correction. Such systems provide a robust clinical service, and the patient dosimetry associated with the provision of attenuation correction data from the radionuclide transmission sources is considerably less than standard multislice CT protocols [14]. However, the introduction of low-resolution CT scanning protocols within hybrid SPECT/CT systems has reduced the effective dose considerably [15]. Most clinical nuclear medicine departments have replaced their radionuclide-based transmission systems with CT-based hybrid units, due mainly to the extended scope of potential clinical practice (i.e., not just cardiac work), higher photon flux, and localization/diagnostic imaging capabilities.

6.8 First-Generation CT-Based Systems

Initial research around the potential use of a combined SPECT/CT system was conducted in San Francisco by Dr. Hasegawa and colleagues [16] and involved the use of high-purity germanium detectors to simultaneously acquire X-rays (40–100 keV) and gamma rays (140 keV). Unfortunately, this system required a total examination

time of between 3 and 4 h and was therefore impractical for routine clinical use. The concept of a single imaging environment, which utilized an X-ray-based detector system, in addition to gamma ray detection, was further refined to create a dedicated environment, which included a single-headed gamma camera and a single-slice CT unit. Acquisition times were quicker than the previous prototype, and this approach provided a true representation of anatomical localization and attenuation correction capabilities. The use of an X-ray source in the form of a CT-based unit provides attenuation coefficient data, which is rich in photon flux (lower image noise), compared to other transmission-based methods and provides an option for higher quality anatomical data [15].

Initial hybrid SPECT/CT gamma cameras utilized a low-end-resolution CT unit, which provides an axial resolution of 10 mm [11], and the X-ray tube and detector configuration are located on the same gantry as the detector heads of the gamma camera (Fig. 6.6). Unlike the initial approach taken by Hasegawa and colleagues [16], the SPECT and CT imaging is not simultaneous, with the CT examination being conducted after the SPECT procedure. A four-slice version of the GE Hawkeye provides an axial resolution of 1 cm but still utilizes the same gantry as the gamma camera detectors.

Due to the physical geometry of the first-generation system, the rotation speed of the CT unit is limited to below 2.8 revolutions per minute [17], utilizing slip-ring technology within the gantry. Such low-resolution images are useful for attenuation correction, spatial localization, and fusion with SPECT data, but cannot substitute high-resolution diagnostic CT [18]. The acquisition time for 40 cm field-of-view coverage is 12 min with a first-generation GE Hawkeye system; however, the acquisition time is quicker for the 4-slice GE Hawkeye system. The approximate patient dose from the CT procedure is also considerably lower (20 times less) than a dedicated CT system [11]. In summary, an entry-level SPECT/CT system offers a gateway to the provision of a fundamental hybrid imaging service and an opportunity to incorporate new techniques within a nuclear medicine department.

The introduction of SPECT/CT has redefined the layout of traditional nuclear medicine departments, with the addition of lead shielding required for the protection of operators during the CT examinations. Figure 6.7 demonstrates a typical setup for a first-generation SPECT/CT system (GE Hawkeye), with the inclusion of a secondary radiation protection barrier for the operator. Although the introduction of a secondary radiation barrier presents new ways of working for the traditional practitioner in nuclear medicine, the acquisition and possible processing workstations are now positioned behind a screen, which is less visible to patients. Certain acquisition operations are now also possible from behind the protective screen (e.g., commencement of examinations), which may minimize personal radiation dose from the patient during conventional nuclear medicine and/or hybrid examinations. However, consideration should be given to the impact upon future working practice of staff within this new environment, particularly when undertaking certain patient examinations [19, 20].

Within multisystem departments, the introduction of a SPECT/CT unit may require reorganization of available space, particularly if an open design is present.

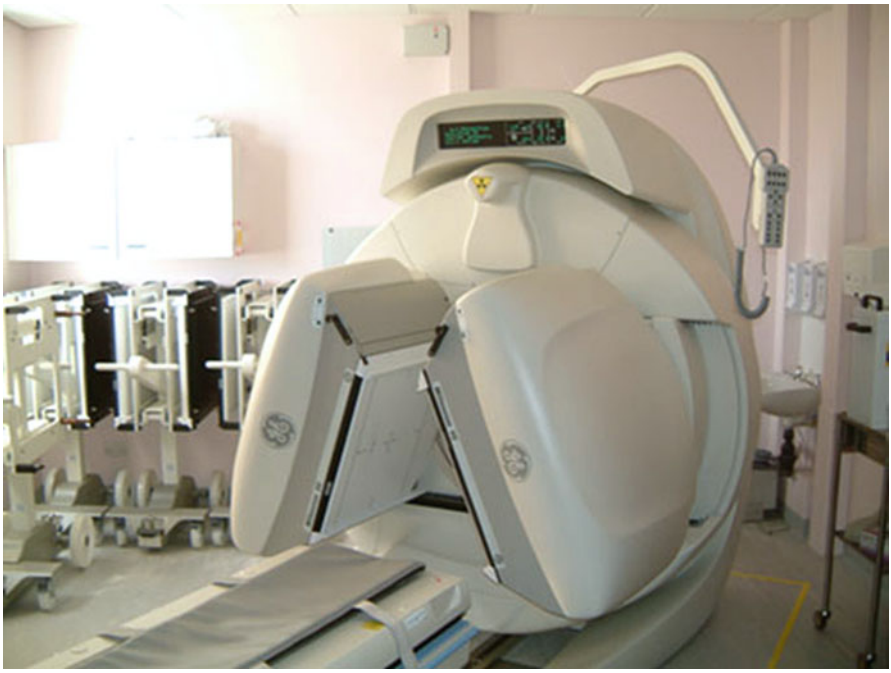

Fig. 6.6 GE Hawkeye SPECT/CT system, with integrated X-ray unit, consisting of a low-output X-ray tube and cadmium tungstate detectors

Fig. 6.7 Shielding area for operator within a SPECT/CT environment

Figure 6.8 demonstrates an example multisystem departmental layout, and the use of retractable/movable lead screens is utilized during SPECT/CT examinations. The SPECT/CT system incorporated within Fig. 6.8 is a GE Hawkeye, which has a low X-ray tube output of between 1 and 2.5 mA. Consideration should however be given to operators working in other parts of an open space department, to ensure they are not at risk of unnecessary exposure during a hybrid imaging examination.

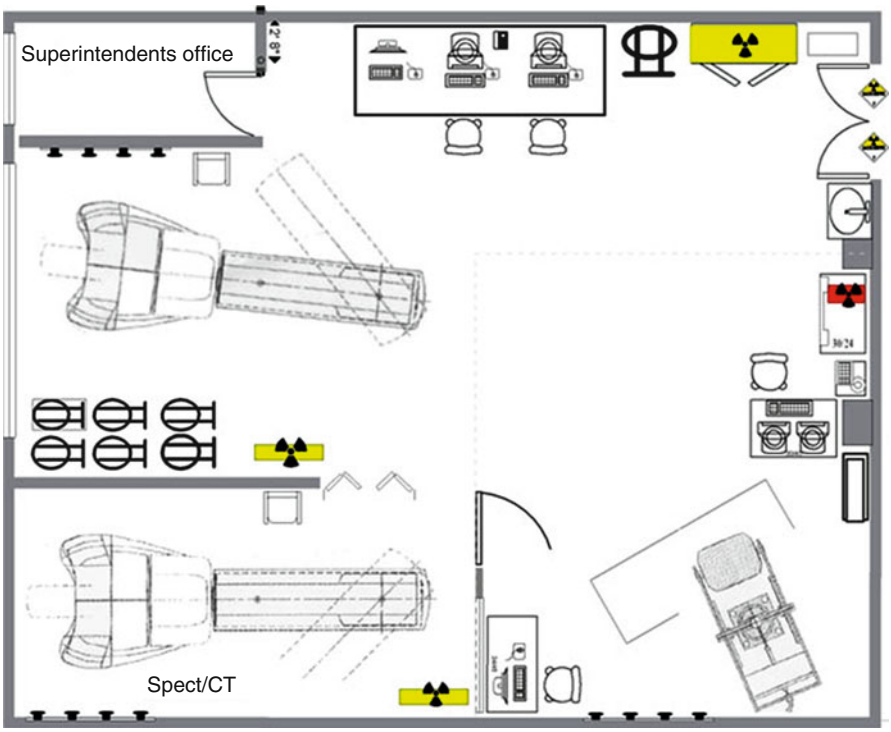

Fig. 6.8 Example layout of a multisystem department, demonstrating configuration of operating areas and available floor space. The space required for a GE Hawkeye and Hawkeye 4 is similar to a conventional gamma camera SPECT system (14 × 16 ft) [11]

6.9 Modern SPECT/CT Systems

Modern hardware engineering developments have advanced at a rapid pace over the last 5 years, and the inclusion of a dedicated high-specification CT unit within an existing gamma camera is beginning to redefine patient treatment and subsequent management [13]. Multislice CT units, coupled to modern gamma camera units, also utilize clinical features such as laser positioning markers and semirigid imaging pallets, to ensure accurate co-registration of emission and transmission data, for the purposes of image fusion applications.

The availability of separate gantry systems (Fig. 6.9) overcomes the relatively inferior resolution and slower CT acquisition speed, which is associated with first-generation SPECT/CT systems. This has however created the need for additional cooling systems for the CT components, which is either air or water controlled and adds increased noise levels within the imaging environment, compared with first-generation systems.

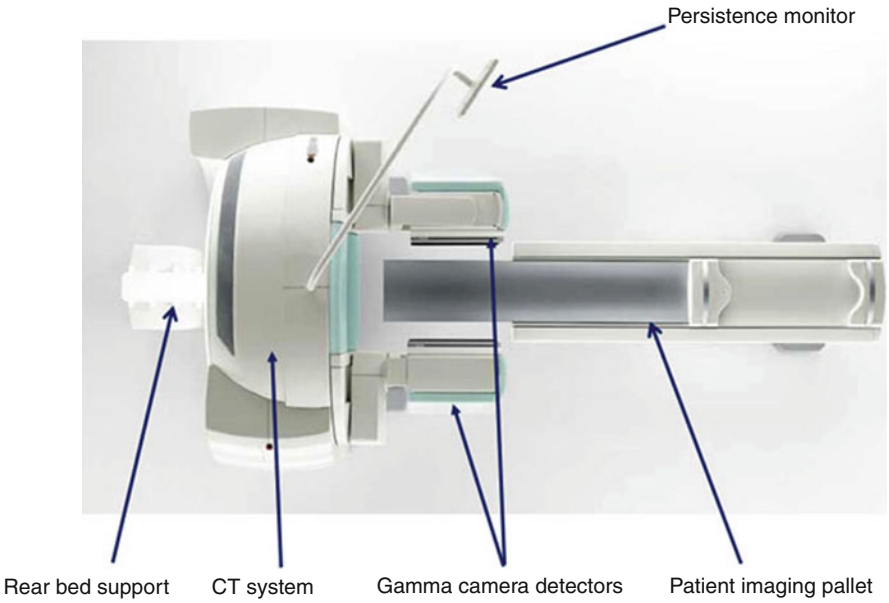

Fig. 6.9 Overhead view of a modern SPECT/CT system, demonstrating the separate imaging components within one physical environment

In addition, greater consideration is now required for the accurate acquisition of CT data, as it will be performed quicker than conventional first-generation systems, which may lead to respiratory motion artifacts. SPECT/CT systems with 64-slice technology are now commercially available, providing an extended scope of clinical examinations, such as cardiac CT and stand-alone, contrast-enhanced CT techniques. The main advantages of multislice CT units include shorter scan times, extended scan ranges, and improved longitudinal resolution capabilities. The operator also has a choice of how the CT data is acquired (i.e., in terms of spatial resolution), with various detector array combinations possible with higher multislice CT units [11].

Recently, the emergence of a SPECT/CT system with a flat panel X-ray detector has been introduced within clinical practice (Fig. 6.10). The Philips BrightView system offers an ability to acquire coplanar and SPECT/CT and utilizes a digital amorphous silicon detector panel, which is retractable. Although the CT unit is on the same gantry as the gamma camera detectors, this system is capable of performing 5 rotations per minute, which is considerably quicker than other systems with a similar geometry design. The floor space required for a multislice SPECT/CT unit is considerably bigger than a first-generation system, with 15×24 ft being documented by some manufacturers. This is considerably larger if you include a separate operator's console as well. The floor loading space should also be

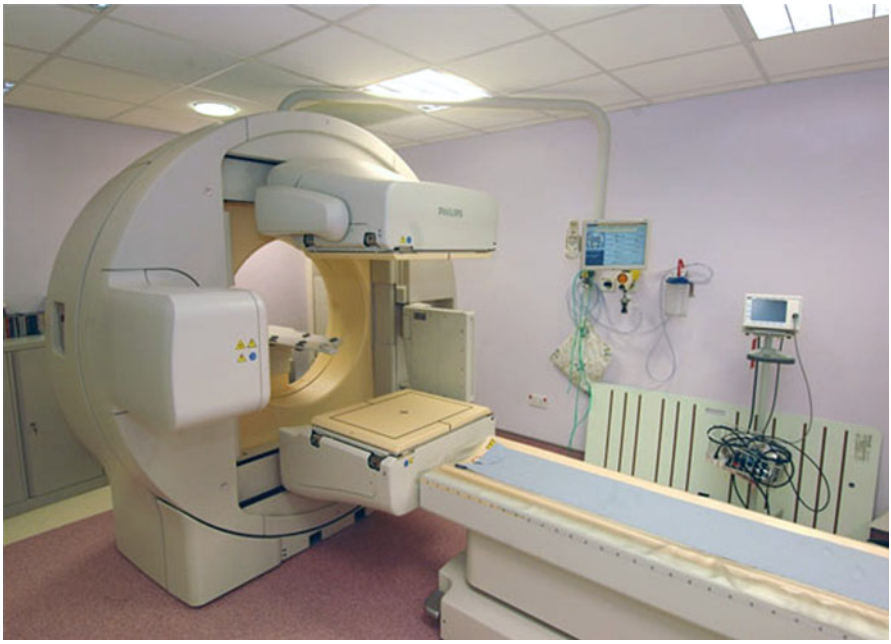

Fig. 6.10 Philips BrightView XCT system, featuring a retractable flat panel X-ray detector system

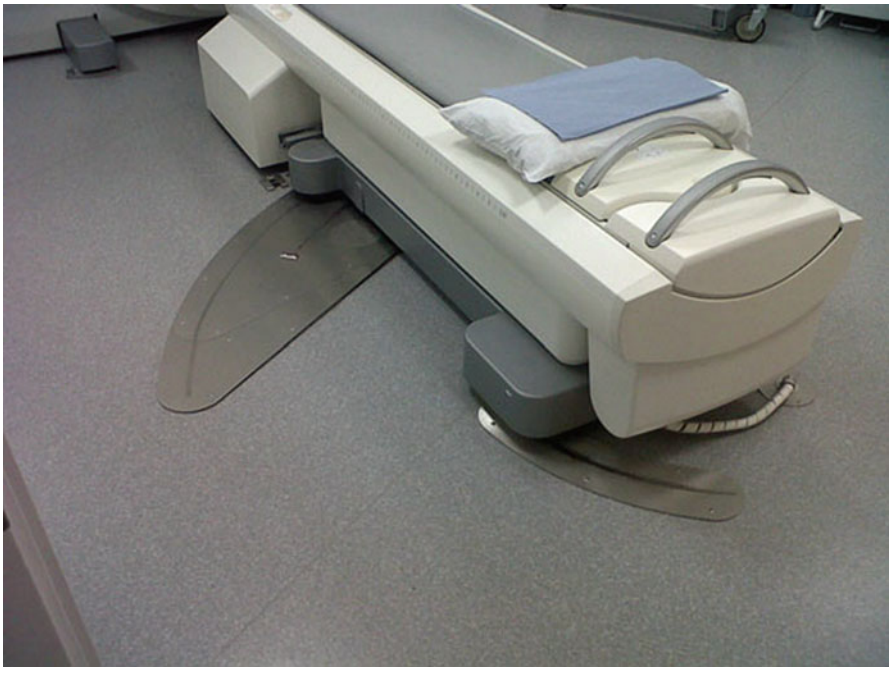

Fig. 6.11 Installation of floor plates to preserve the alignment of the patient imaging pallet to the gantry

assessed, as the weight of a multislice SPECT/CT system is considerably heavier than a GE Hawkeye unit, almost double in some instances [11]. In some departments, the installation of floor plates (Fig. 6.11) has been undertaken to ensure the registration between the patient imaging system and gantry is accurate over time.

6.10 Additional Considerations for Introducing a New SPECT/CT System

6.10.1 Patient Weight Capacity of Patient Imaging Pallet (Bed)

Consideration for the maximum weight of the imaging pallet needs to be factored, especially if the patient is going to have a CT examination. Although all SPECT/CT systems have a rear bed support (Fig. 6.8), there is potential for *image sag* between the emission and transmission data sets with obese patients. Most SPECT/CT systems will guarantee weights up to 200 kg (440 lb), with a pallet deflection (image sag) of less than 2 mm at 90 kg.

6.10.2 Minimum Height of the Patient Imaging Pallet (Bed)

Ideally, the minimal height of the patient imaging pallet should be low enough to accommodate wheelchair patients but also high enough to transfer patients safely from a bed/stretcher.

6.10.3 Travel Length of the Patient Imaging Pallet (Bed)

Typically, this will be in the region of 200 cm, to allow multiple bed positions to be fused together within SPECT/CT examinations. First-generation SPECT/CT systems had limited z-axis coverage, due to the design of the hybrid imaging unit. However, the introduction of integral bed and CT gantry designs has created additional coverage and flexibility for multi-bed SPECT/CT examinations.

6.10.4 Auto Contour and Positioning Optimization

All modern SPECT/CT systems should offer an auto contouring facility for SPECT examinations and the ability to plan the CT examination from the SPECT data (often referred to as “guided CT”).

6.10.5 Automated Collimator Configuration

Having a wide range of collimators to accompany the purchase of a new SPECT/CT system may be inevitable, especially if the range of examinations increases due to the availability of new imaging procedures (e.g., In111, I131). Some SPECT/CT systems offer a semiautomated form of collimator exchange process, which reduces the amount of manual handling associated with this task. However, the positioning of collimator carts within a bust nuclear medicine department needs considerations, especially if they contain high energy collimators.

6.10.6 Type of CT Unit

Some manufacturers offer a range of CT units for use within a SPECT/CT system. Typical entry-level SPECT/CT systems will provide CT data with an axial resolution of between 3 and 5 mm, whereas higher specification units will generate an isotropic resolution (e.g., 0.33, 0.63 mm) output.

6.10.7 CT Tube Loading

Initial SPECT/CT systems only offered tube loadings in the region of 10 mA; however, modern mutlislice CT systems offer tube loadings of between 30 and 240 mA, with higher specification systems providing up to 345 mA.

6.10.8 Dose Modulation

All modern (SPECT/)CT systems will incorporate dose modulation, in order to minimize patient dose from the CT examination.

6.10.9 Laser Positioning Lights

Laser lights aid in the positioning of patients for hybrid and stand-alone CT procedures.

6.10.10 Automated Routine Quality Control Mode

Some systems now offer the ability for automated routine/daily quality control checks, with an integrated point source. This permits the daily quality control checks to be performed prior to the commencement of the working day.

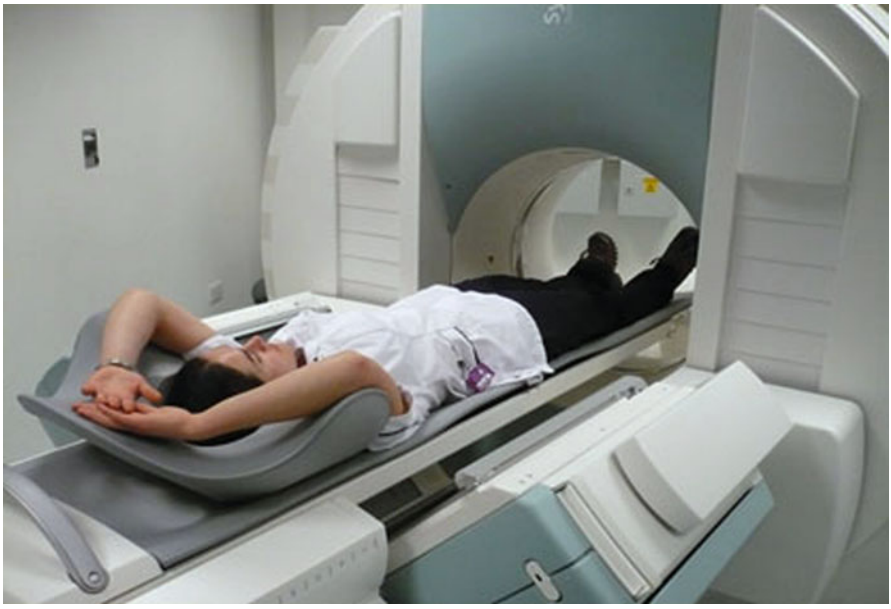

Fig. 6.12 Variable geometry of gamma camera heads on a modern SPECT/CT system, reducing the "closed" space during the CT procedure

6.10.11 Size of CT Patient Bore

Consideration for the size of the CT aperture is needed, in order to identify any potential truncation artifacts from large patients undergoing a SPECT/CT procedure. Typical gantry aperture sizes within SPECT/CT systems are in the region of 70 cm, with a useful scan field of 50 cm.

6.10.12 Degree of Flexibility with the Gamma Camera Detectors

The greater the degree of flexibility with the gamma camera heads, the potential for patient compliance will become apparent. In addition, some equipment manufacturers position the gamma camera detectors in to the "L" mode (i.e., cardiac imaging) underneath the patient bed during the CT acquisition, in order to provide the appearance of open space for the patient (Fig. 6.12).

6.10.13 Integration of Flat Panel Imaging Pallet

The ability to fix a flat carbon fiber pallet to an existing patient imaging system provides an imaging environment which offers the capability to acquire data for radiotherapy planning and/or dosimetry purposes.

6.10.14 Resolution Recovery Hardware/Software

Manufacturers now offer a new generation of focused collimators which offer the prospect of reduced scanning time for certain examinations. Coupled with advanced resolution reconstruction software algorithms, clinical departments are revisiting their traditional patient workflow patterns and introducing new accompanying imaging techniques, such as calcium scoring.

6.10.15 Flexibility of CT Acquisition Parameters

Modern SPECT/CT systems offer the operator the ability to acquire the CT data at various levels of resolution/quality.

6.10.16 CT Processing Parameters

Equally, there are also a number of processing parameters associated with the acquired CT data, which can be employed by the operator.

6.10.17 Integrated ECG Hardware Port and Output Display

The ability to acquire an ECG signal during cardiac examinations should be relatively straightforward. An integrated ECG port on the SPECT/CT unit (Fig. 6.13) and a display on the persistence scope provide the operator with flexible working environment.

6.10.18 Communication and Patient Monitoring Aids

A voice intercom and patient monitoring camera can provide additional monitoring during the CT examination. In addition, the use of visible lights for patients can provide an indication when the CT examination is about to begin.

6.10.19 Patient Positioning Supports

The use of a dedicated head support during neurological examination provides support/immobilization for the patient, close contact for the gamma camera detector heads, and an isocentric point of reference for the CT examination, if being performed.

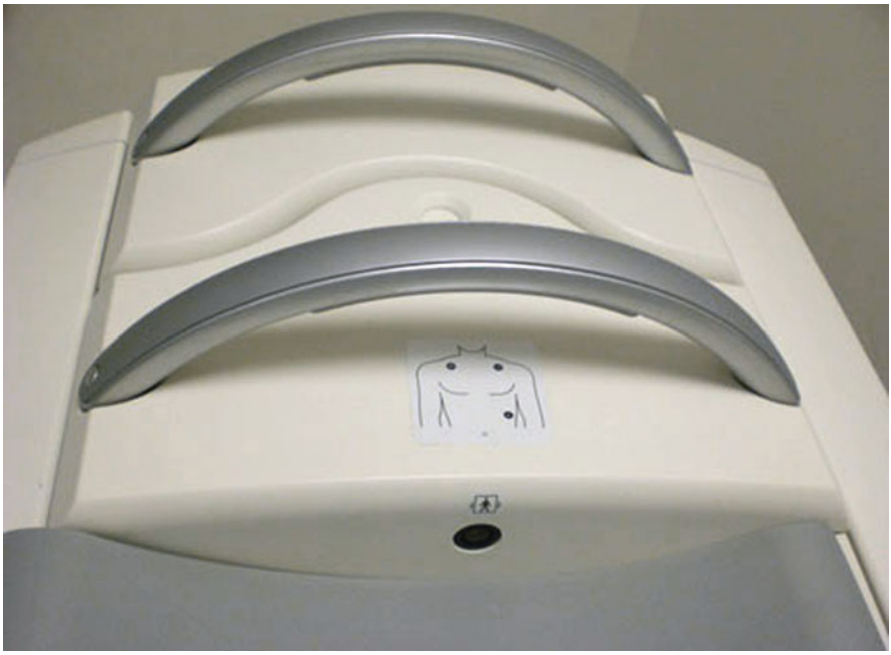

Fig. 6.13 Integrated ECG port on the patient imaging pallet allows seamless acquisition of cardiac gating data during myocardial perfusion examinations

Most manufacturers adopt the use of carbon fiber within the construction of imaging beds and head supports. The use of carbon fiber represents a low atomic number material, which will absorb less than 10 % of gamma energy (at 140 keV).

6.10.20 Environmental Noise

The use of a multislice CT unit requires either additional air or a water cooler in order to maintain an operating temperature for the CT components.

6.11 Conclusion

The financial cost of implementing a SPECT/CT hybrid imaging system needs to be assessed early on in the procurement process as possible. Additional work will need to be performed on your existing examination room (e.g., addition of a lead glass screen/separate operator's console) and ensuring there is enough space for the unit. In addition, the maintenance contract of the hybrid imaging unit may be slightly higher than a conventional gamma camera, to include the CT unit [19].

6.12 Software Considerations for SPECT/CT

6.12.1 Acquisition Software

SPECT/CT systems have acquisition setup software that allows various parameters to be specified for both nuclear medicine and CT imaging. For SPECT/CT specifically, there are several options for defining how imaging is carried out, and these need to be carefully considered when purchasing a system.

Acquisition protocols can typically be set up to acquire either the SPECT followed by the CT or vice versa. In the first case, systems may then allow the scope of the CT to be determined from the SPECT images. This would be useful, for example, if a SPECT scan had identified a small focus of activity which would benefit from being fused with a CT image; the CT field of view could then be defined to encompass just the extent of the activity, thereby reducing CT exposure to the patient. This is termed SPECT-guided CT.

Alternatively, it may be possible to acquire the CT first. In this instance, a topogram or scout view would typically be acquired, from which the area to perform CT on is determined. This would then be followed by a SPECT image covering the same volume.

In terms of reducing patient exposure, some systems can perform a low-dose CT scan for the purposes of attenuation correction and SPECT/CT fusion, and this may be an advantage if fully diagnostic CT is not required. The availability of diagnostic quality CT on a SPECT/CT system may, however, be beneficial in obviating the need to carry out a separate CT scan if clinically indicated. These systems should also be capable of carrying out low-dose CT for the purposes of attenuation correction, for which high-resolution CT is not necessary. Most SPECT/CT systems will have automatic dose-reduction algorithms, which should be evaluated during the tendering process.

6.12.2 Processing Software

Processing software will include tomographic reconstruction of the SPECT and CT data.

Options for SPECT reconstruction are likely to include filtered back projection (FBP), with a range of suitable filters, and iterative techniques such as ordered subset expectation maximization (OSEM). Processing systems should allow different types of pre- and post-filters and the ability to vary the parameters to control the extent and magnitude of smoothing before and/or after reconstruction. Most systems also allow some form of resolution recovery, which can correct for the lost of spatial resolution with distance from the gamma camera face and is purported to give equivalent image quality with a reduced number of counts. This allows reduction of imaging time and/or activity (and hence radiation dose to the patient). Scatter correction is also generally an option within SPECT reconstruction for modern-day systems.

Correction for the attenuation of gamma rays within the body is important in some applications and comes as standard on most SPECT systems. Attenuation correction takes into account the differing amounts of attenuation of gamma rays as they traverse the body. This attenuation will vary with projection angle, as different anatomical structures will obstruct the gamma rays depending on the direction of the path between their origin and the gamma camera. Attenuation correction can be performed using simple geometric assumptions, for example, a uniform cylinder, but the availability of the CT on a SPECT/CT system allows this to be performed much more accurately since the CT images give an accurate anatomical map.

The accuracy of attenuation correction depends on having proper registration between the two modalities. This is essential as nonaligned images will lead to errors in the calculated attenuation factors which, in turn, can cause artifacts in the reconstructed images. The system display should allow corrected and non-corrected images to be displayed side by side to indicate the presence of such artifacts. As the SPECT and CT procedures are not performed simultaneously or in the same bed position, misregistration may occur for two reasons: (i) a systematic error in the positioning offset between the SPECT and the CT and (ii) patient movement between the SPECT and CT imaging procedures. The former is calibrated at installation and should be checked periodically (see Sect. 4.5). The latter will be somewhat patient dependent but may be reduced through adequate explanation of the process to the patient beforehand and proper patient supervision throughout the whole imaging process. Although (i) may not be an issue, it is essential that the processing software allows manual registration of the SPECT and CT data sets to overcome situations where the patient has moved between studies. The ease of use of such software should be evaluated as part of the tendering process.

Properly aligned SPECT and CT data sets can then be used to allow accurate attenuation correction. Typically, software will calculate an attenuation map from the CT slices. This is a set of slices of the same resolution as the SPECT data with values representing attenuation coefficients that are derived from the CT number. The attenuation map is then used in the iterative reconstruction algorithm. Normally, this is an automatic part of the processing with little or no control other than to switch it on or off.

SPECT/CT image fusion display is another important part of the processing software. Typically, this will allow SPECT and CT images to be overlaid. It is useful if the software allows control of the blending or weighting between the two data sets from one extreme, that is, 100 % CT 0 % SPECT, and to the other, that is, 0 % CT 100 % SPECT. Color scales or lookup tables (LUTs) are also very important when trying to assess SPECT/CT fused images. CT is typically displayed in a linear grey scale, whereas the SPECT data is displayed in color. Figure 6.14 shows a SPECT-CT fusion display with weightings: (a) 100 % CT, (b) 70 % CT 30 % NM, (c) 50 % CT 50 % NM, and (d) 70 % CT 30 % NM. Some systems allow other display options, such as count contours, which can also be useful in certain situations, as shown in Fig. 6.14d.

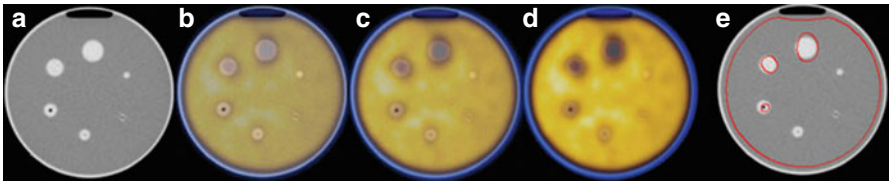

Fig. 6.14 (**a**–**e**) SPECT-CT fusion display with weightings: (**a**) 100 % CT, (**b**) 70 % CT 30 % NM, (**c**) 50 % CT 50 % NM, and (**d**) 70 % CT 30 % NM (**e**) Count contour type display

Tools, such as triangulation cursors and slice scrolling, are generally available to simplify localization of lesions. Some systems may also offer 3D image fusion, which may be useful in certain situations.

Archiving of images is another area of software that needs to be considered. This can be done via local backup media such as optical disks or DVDs and/or through a picture archiving and communication system (PACS). The latter allows images to be stored on a central server, together with other images from other modalities and other data such as clinical reports and medical photographs. Images stored on PACS are typically in Digital Imaging and Communications in Medicine (DICOM) format, which is a medical imaging standard allowing images to be transferred between and viewed by any DICOM-compliant software. Options such as automatic archiving, for example, overnight, and the types of images and data that can be stored, should be carefully considered.

6.13 Quality Control Tests for SPECT/CT

Aside from the usual range of quality control tests that are needed to maintain the separate gamma camera and CT elements of the SPECT-CT scanner, the combination requires additional testing in two respects – registration and accuracy of attenuation correction.

6.14 Registration, or Alignment, Between the Two Modalities

Precise registration is essential in achieving accurate attenuation correction and SPECT/CT image fusion. There are two elements that affect the accuracy of SPECT-CT registration:

(a) *Patient movement between the SPECT and CT imaging procedures*
This includes not only change of position of the patient between scans but also involuntary movement due to respiration, cardiac motion, etc. Quality control on a per-patient basis must be carried out using appropriate system software that

Fig. 6.15 SPECT-CT alignment phantom

allows clear assessment of SPECT/CT registration and the ability to perform a manual reregistration if necessary.

(b) *The mechanical offset between the SPECT and CT gantry positions*
This is calibrated by the manufacturer's engineers at installation of the system, thereby providing an automatic means of correction thereafter. This needs to be checked on a periodic basis as part of the system's routine quality control and immediately following any gantry adjustments. The frequency of testing depends on the stability of the system, and the European Association of Nuclear Medicine (EANM) Physics Committee have proposed that this be done at least monthly [21]. This is achieved by using a test object which has structures that are visible on both SPECT and CT; for example, the phantom shown in Fig. 6.15 could be used. This has hollow spheres which can be filled with a solution of Tc99m, allowing them to be visualized on the gamma camera as well as the CT.

A SPECT/CT scan is then carried out and the image fusion software is used to ensure accurate alignment. Figure 6.16 shows some of the resulting fused images for a system with good registration between the two modalities. The colored nuclear medicine images clearly show uptake in the sphere and cylinder volumes, the boundaries of which are highlighted with the CT.

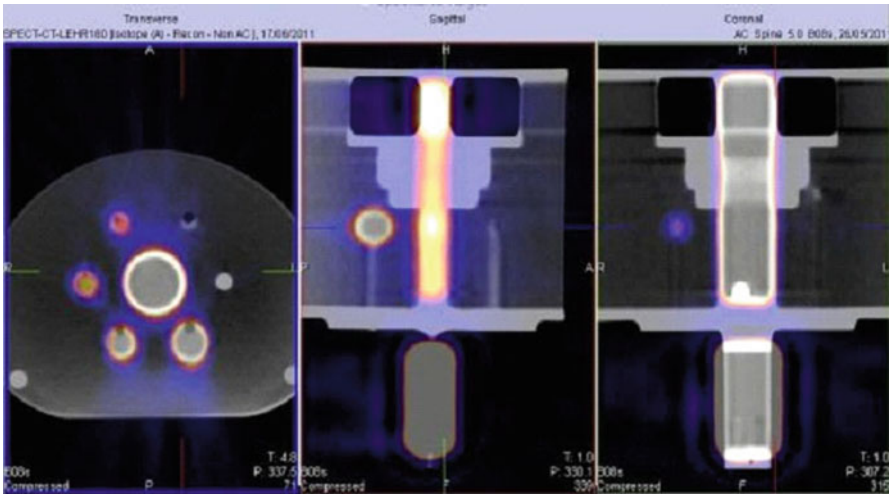

Fig. 6.16 Example of good SPECT-CT registration

Fig. 6.17 SPECT-CT alignment check setup for GE Hawkeye

Apart from such generic phantoms, manufacturers generally supply specialized test objects for the purpose of testing SPECT/CT alignment, which can also be used to recalibrate the offsets should it be necessary. These test objects vary from one

manufacturer to another; some use an array of small gadolinium-153 (Gd-153) disks mounted on a frame. The Gd-153 sources emit gamma rays that are detected and imaged by the gamma camera and that are dense enough to show on the CT images. Others use syringes filled with Tc99m solution mounted in a foam structure (as shown in Fig. 6.17) or small vials which are filled with a Tc99m solution mixed with CT contrast media.

There are advantages and disadvantages to these methods which need to be considered when purchasing a system. For example, test objects that utilize long-lived, sealed sources, such as Gd-153, will need to have those sources replaced several times during the lifetime of the camera, since the half-life of this radionuclide is around 8 months. The resupply and radioactive waste disposal of these sources will incur a cost each time this is done, and this needs to be borne in mind when costing the system. Conversely, although the methods that rely on containers that are filled with Tc99m solutions do not have that disadvantage, they do require significantly more time on each occasion to set up and use. It is largely then a balance between cost and convenience.

6.15 Accuracy of Attenuation Correction (AC)

Proper correction of attenuation will produce images with accurate reproduction of the radionuclide distribution. This is especially important when quantification of uptake is required. AC is commonly used in myocardial perfusion imaging where it can correct for differences in attenuation, which is particularly problematic in the anterior wall in women with large breasts and in the inferior wall for obese men. Accuracy of AC depends on SPECT-CT registration (see (i)) and on the translation of CT data to an attenuation map. This latter aspect relies on attenuation coefficients derived from CT numbers.

Accuracy of AC is not normally assessed by manufacturers at installation, and test objects are not normally supplied to enable this to be tested. Third-party test objects, however, are available, such as those in Fig. 6.15. This phantom has 2 refillable cylindrical chambers of the same diameter, one outside the phantom and the other inside. Both can be filled with a Tc99m solution. The activity in the inside phantom is subject to variable attenuation from water, liquid bone, and air. A SPECT-CT acquisition is performed and the SPECT data attenuation corrected. A profile drawn through the center of the cylinders on a sagittal or coronal slice will indicate whether AC is correcting properly.

Figure 6.18 shows example profiles of non-AC vs. AC overlaid on a sagittal CT slice to show the structures within the phantom. In this example, the AC data is a significant improvement over the non-AC but has overcorrected by about 20 %.

It is quite possible to test AC using locally constructed test objects, for example, using Tc99m-filled syringes inside and outside different attenuating media.

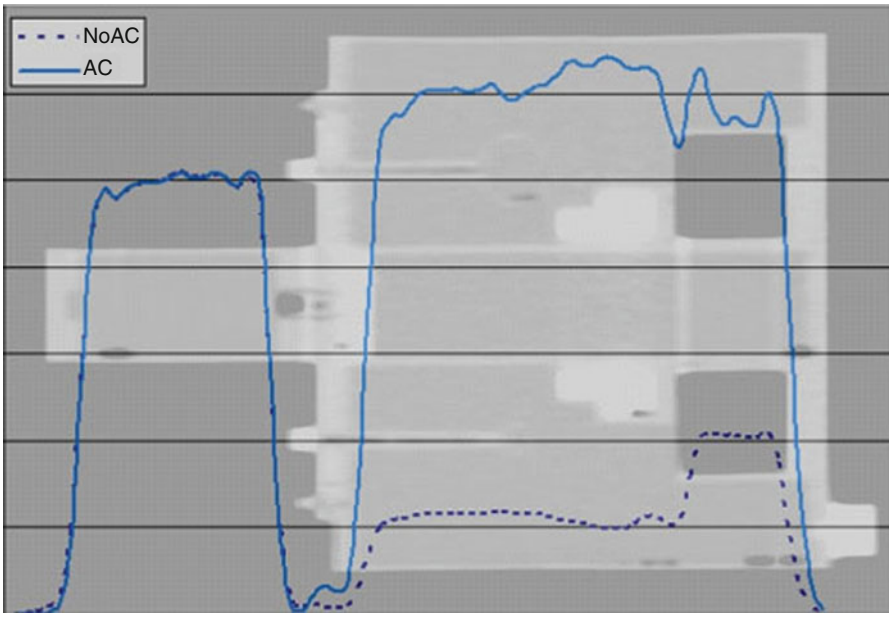

Fig. 6.18 Assessment of attenuation correction results

6.16 Emerging Techniques for SPECT/CT

The emergence of SPECT/CT systems within the nuclear medicine environment has provided the opportunity to perform new techniques and interprofessional working opportunities [22, 23]. Modern multislice computed tomography (MSCT) technology coupled with state-of-the-art photon detection systems has provided a realistic solution to attenuation and registration issues [24]. Hybrid SPECT/CT systems are now accessible in most nuclear medicine departments, along with associated sophisticated software to correct for the effects of misregistration and motion artifacts.

The integration and clinical use of CT within a SPECT imaging environment has been slower than PET/CT, but over the last 10 years, this balance has been addressed, with modern multislice SPECT/CT units now having the diagnostic capability to provide a high level of anatomical information, which is considered important for the management of patient pathways [25]. Over several years the reemergence of established nuclear medicine techniques, involving radioisotopes such as gallium (Ga67) citrate, indium (In111) octreotide, iodine (I131), and thallium (Tl-201) chloride, has been documented [26–29]. This has led to a steady resurgence of SPECT/CT techniques within the nuclear medicine department [25] and a greater level of diagnostic confidence and overall involvement in the patient management.

Clinical areas where SPECT/CT is emerging as a "front-line" technique include: Benign/malignant skeletal tumors: The use of SPECT/CT minimizes the number of equivocal findings normally associated with planar imaging and is beginning to be

utilized as a whole body protocol to characterize indeterminate lesions in patients with known malignancies [30]. Sentinel node lymphoscintigraphy: The use of SPECT/CT for sentinel lymph node mapping is an area, which has brought together health-care professionals and begun to transform clinical practice. SPECT/CT techniques improve the accuracy of anatomical localization of the sentinel nodes, especially if they are close to the injection site [31]. Preoperative assessment of parathyroid adenomas: The use of SPECT/CT increases the overall sensitivity and specificity of parathyroid adenomas and also aids in the localization of potential ectopic and/or multiple adenomas [32]. The inclusion of nuclear medicine practitioners in the preoperative assessment of parathyroid adenomas, working in conjunction with other professional groups, has a positive impact on workforce dynamics. Neuroendocrine tumors: The value of SPECT/CT and the use of radiopharmaceuticals such as In111 octreotide in the detection of neuroendocrine tumors (NETs) are well documented [33–35]. The use of SPECT/CT as part of the imaging technique for the detection of NETs has been reported to change the overall management of patients in over one-third of cases [36] and impact on the type of therapeutic treatment [37]. Figure 6.19 provides an example of the clinical value of In111 octreotide SPECT/CT imaging.

Selective internal radiation therapy (SIRT): The use of SPECT/CT can be useful in the pre- and post-therapy imaging of SIRT techniques, such as yttrium-90, which is used for the treatment of non-resectable liver tumors. Integration of coronary angiography and calcium scoring CT: The emergence of multislice diagnostic CT units with low-dose acquisition protocols and prospective ECG triggering offers a realistic opportunity to perform hybrid imaging techniques on different risk groups of patients [38]. Using the latest iterative reconstruction algorithms for SPECT myocardial perfusion imaging (and a reduced radioactivity dose) and a low-dose CTA results in the production of hybrid images with a radiation dose of less than 3 mSv [39].

6.17 Changes to Working Practice

Referrals for examinations requiring new isotopes/radiopharmaceuticals may require modifications to various aspects of working practice, including the development of new clinical protocols, which may encompass anatomical-based imaging. Combined SPECT/CT procedures should be chosen on an individual basis, justified, and reflect clinical need [40]. Decisions should be made related to the appropriate use of CT for the specific examination being performed. The use of diagnostic CT within a SPECT/CT examination may increase the overall patient dose by up to 14 mSv [41]. Use of low-dose CT may reduce the additional patient dose to a range of 1–4 mSv [40] and may be useful for attenuation correction or localization purposes.

Modern technology and resolution-based software methods have provided greater access to hybrid imaging techniques within most nuclear medicine departments. Such developments have also placed greater responsibility on the practitioner to engage with new imaging and processing protocols and possible use of ionic contrast media

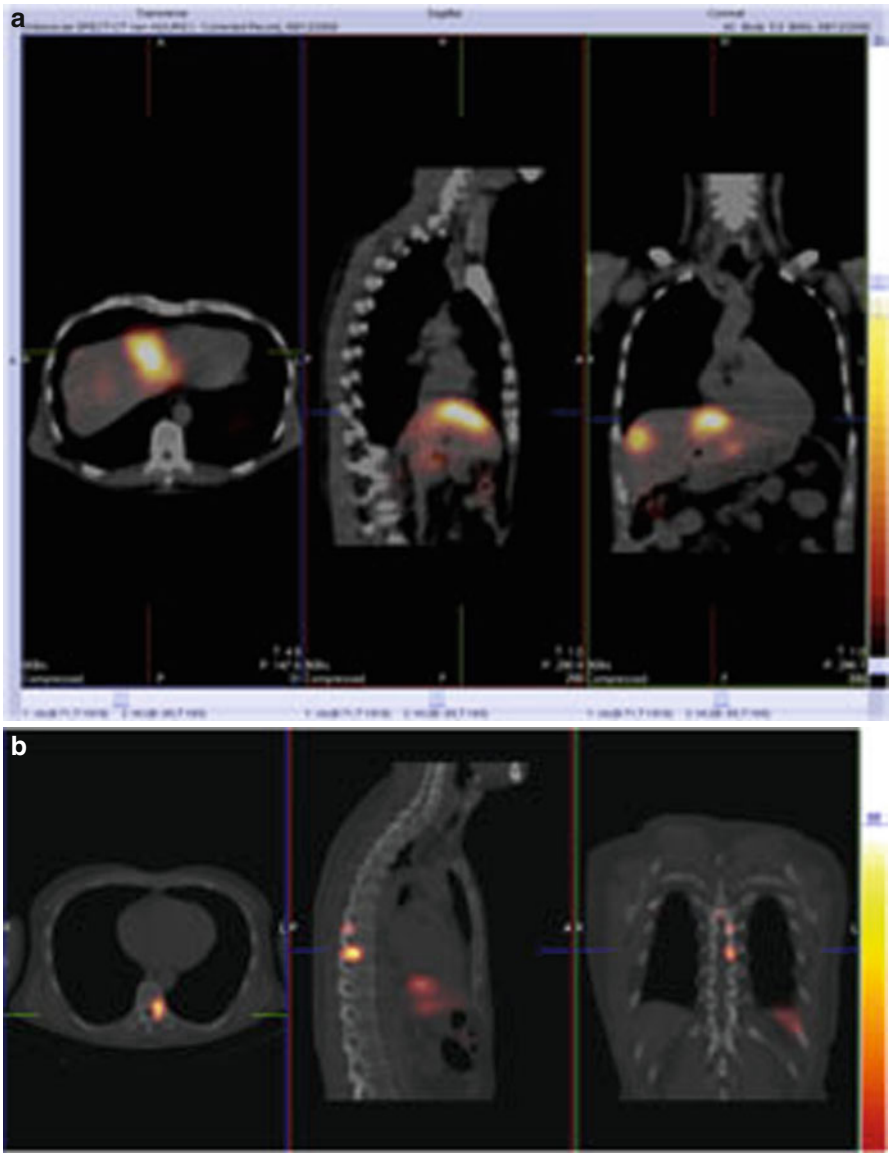

Fig. 6.19 (**a**) SPECT/CT of patient with known diagnosis of metastatic carcinoid, with mixed response to treatment. (**b**) In111 octreotide examination performed to explore the possibility of radiolabelled octreotide treatment. Octreotide uptake was demonstrated within the liver and metastases within the skeleton

examinations. Equipment manufacturers are also responding to the changing nature of nuclear medicine practice, by offering flat imaging couches as part of a tender response, in order to accommodate possible radiotherapy planning examinations (Fig. 6.20).

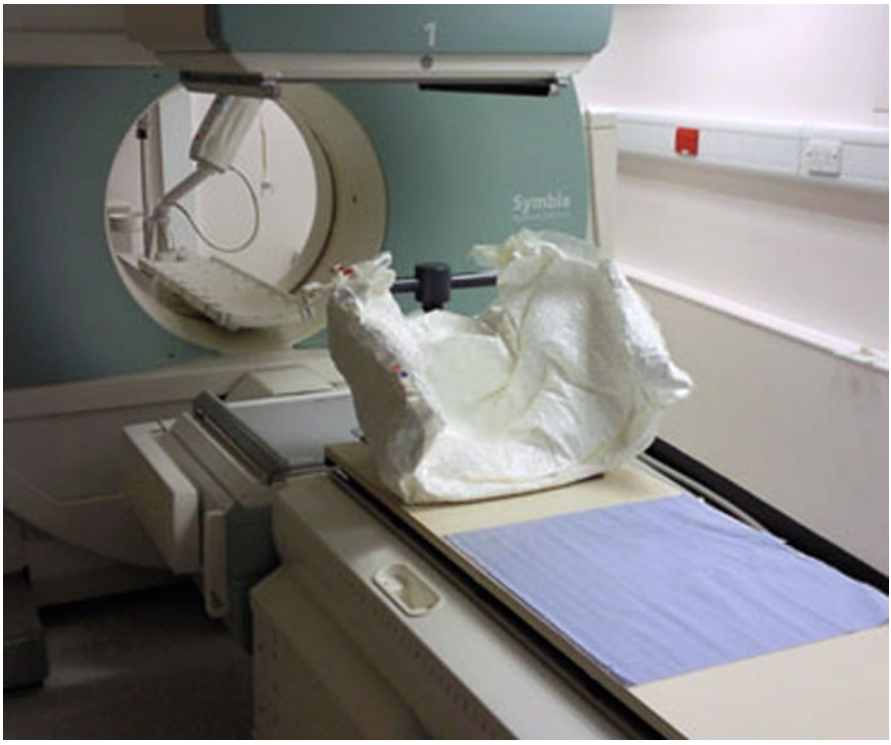

Fig. 6.20 Adaptation to conventional nuclear medicine environment. Inclusion of a flat imaging couch and stereotactic frame for radiotherapy patients

Clinical departments are beginning to explore the value of performing dosimetry-based examinations within a SPECT/CT environment [42] with a view to providing valuable planning information prior to external beam radiotherapy treatment [43]. Using CT as part of a nuclear medicine examination is useful for tumor staging, restaging, and the possible identification of pulmonary nodules, which may show as negative on functional images [40].

The monitoring of patients' post-ablation therapy for the treatment of thyroid carcinoma has provided an opportunity for skill mix and greater involvement in the overall patient management. Traditional planar imaging of I131 is characterized by poor anatomical representation [44]. The use of SPECT/CT for the localization of differentiated thyroid cancer improves the overall diagnostic accuracy and provides a use for nuclear medicine as a monitoring tool, leading to the modification of ongoing treatment as high as between 35 and 47 % in some patients [45].

The involvement of SPECT/CT within the patient's overall medical management also impacts on the unnecessary need for invasive surgery [44] and provides new prognostic imaging techniques such as calcium scoring and multiplexing imaging approaches, such as SPECT V/Q and CTPA examinations. Terms such as "one-stop shop" are beginning to emerge whereby patients, especially those who live a

geographical distance from their hospital, may have a diagnostic CT examination as part of a joint hybrid imaging procedure. This may provide additional benefit to the patient's management, particularly where incidental findings are identified.

The clinical introduction of solid-state semiconductor materials, such as cadmium zinc telluride, continues to improve the sensitivity and intrinsic resolution compared to conventional-based designs. This will provide an opportunity to use hybrid imaging technology within compact footprints. Departments are however advised to consider the additional time that may need to be factored in when delivering new imaging services [23]. Optimization of patient workflow, processing time, and careful consideration of the potential increase in staff personal dose levels from the increased use of longer-lived radioisotopes should be explored when introducing new techniques within the nuclear medicine department.

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Chapter 7
Design of a Suitable Facility to House a SPECT/CT Scanner

David Wyn Jones, Julian MacDonald, and Peter Hiles

7.1 Introduction

On occasion nuclear medicine has been an afterthought where hospital planning is concerned, with genuinely well-designed facilities catering for the needs of staff and patients, capable of housing complex imaging equipment, being addressed in a suboptimal fashion. Of those departments who have been fortunate enough to design their own facilities, until quite recently, they have not needed to consider the additional requirements of SPECT/CT (and PET/CT) from a planning point of view. Not only does the additional feature of CT require more specific radiation protection measures compared to conventional nuclear medicine, as discussed later in this chapter, but consideration also needs to be given to the nature of its clinical utility.

The imaging flexibility offered by HP-CT SPECT/CT systems in allowing occasional use as a conventional CT imaging tool makes it desirable at the planning stage of installation, to consider location in relation to other facilities (e.g., accident and emergency or other CT suites) as well as more conventional facilities used for nuclear medicine.

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7.2 The Scanning Room

The room should have sufficient dimensions to facilitate the largest footprint of any system currently available on the market, so as not to limit the choice of scanner at the procurement stage. Those less fortunate will be left with deciding what equipment will fit into the available space; alternatively, there will be a need to engage architectural opinion to determine whether extra space can be created through extension or reworking of existing space. Examining other options for locations can be beneficial, as relocation of the nuclear medicine department could offer many benefits, not least in terms of available space but also location of the SPECT/CT unit itself.

Since the patient on the scanning couch will be moved in and out of the CT gantry, space must be made available for this and for unimpeded passage around the extremities of the couch. There should be sufficient space for the transfer of a patient from a bed, trolley, or wheelchair to the scanner couch. The use of a patient slide or a hoist may be required to transfer the patient from the trolley to the scanner couch.

An ideal SPECT/CT scanning room should benefit from having a control room with full view of the patient while they are lying on the imaging table (Fig. 7.1). From the control room, the operator would be able to see the full length of the imaging table as it passes through the CT gantry. This will then enable the operator to see:

- Any body part or item of clothing which might become trapped by the moving parts of the scanner.
- A patient that might not be adequately immobilized or a patient who begins to move during image acquisition.
- A patient who may become unwell during any part of the imaging process. This is particularly important for interventional procedures which might involve remote intravenous injections of radiographic contrast media.

Due to the use of X-radiation, it is also necessary to have a good view of any access routes into the scanning room from the control room; these include the main patient entry door and staff entry points. It is considered good practice for the operator to check for any staff or patients entering the area prior to making an X-ray exposure. In particular, CT exposures can take considerably longer to perform than conventional radiographic exposures; this necessitates the need for a particularly good view of all points of entry from the control room. The operator needs to be vigilant at all times.

Sometimes it is not possible to position the SPECT/CT system and operator viewing window in an optimal position, consequently not all entry points can be viewed easily. In such cases additional safety items or measures can be implemented to minimize risks where a compromise in siting the scanner is made. For instance, the use of a closed circuit TV (CCTV) system can improve the operator's view of the patient from the control room; CCTV can also show entry routes into the scanning area. Of course minimizing the number of entry points into the scanning room is also an option, but this is not always possible. Entry doors not visible from the control room can also have either electronic devices for auto-locking the door during X-ray exposures; manual locks with key codes can also be used for staff entrances.

Fig. 7.1 View from control room window – looking along the imaging table into the gantry. This view allows the operator to have excellent conditions for assessing the status of their patient during imaging

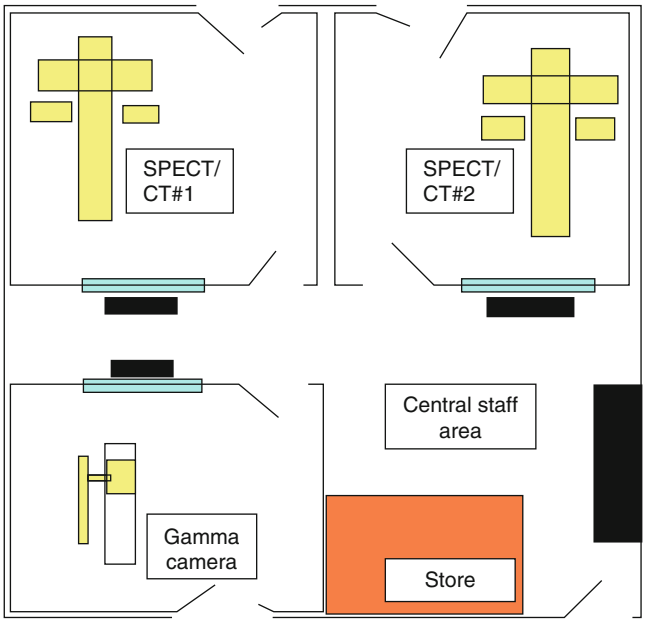

Fig. 7.2 Diagram illustrating central control room shared between scanning rooms

For nuclear medicine departments that have more than one SPECT/CT system, a useful design possibility is to have one (shared) central control room which serves SPECT/CT systems and possibly a conventional gamma camera room (Fig. 7.2). From the shared control room, staff should be able to work together and communicate more effectively. This can have additional benefits in relation to patient care, especially for

Fig. 7.3 This figure demonstrates the use of plastic kickboards which serve to protect the wall

observing patients during image acquisitions, and a second opinion on a given technique is more readily available from nearby colleagues. In addition, should there be an emergency it would be more likely for other staff to be in the vicinity. The use of shared control rooms can therefore result in more effective use of space and possibly also maximize staffing efficiency within the area. If organized well, IV radiographic contrast agents and consumables can be stored centrally in shared control rooms, allowing for easier access and reduced costs of purchasing multiple contrast agent warmers.

Collimator storage is sometimes overlooked when sizing a room for a new scanner. Compromising on collimator storage runs the risk of sustaining damage to these valuable and delicate accessories. Storing collimators in corridors, cupboards, and under staircases remote from the department is not advisable. A particular consideration for nuclear medicine departments is the maintenance of paintwork décor, to aid radioactive decontamination should a spillage or spraying incident occur. Some manufacturer's collimator trolleys can severely damage plasterwork or woodwork in this regard with years of use. Using relatively cheap plastic, covered "kickboards" can be a good design feature (Fig. 7.3).

7.3 Climate Control and X-Ray System Cooling

Conventional gamma cameras have always required air conditioning units to maintain a steady temperature to ensure the stability of the imaging system's electronics and to prevent thermal shock to the delicate sodium iodide crystals inherent in each camera head.

Fig. 7.4 High-capacity climate-control plant installed in a scanning room

The recent addition of HP-CT units to SPECT-CT systems usually requires additional capacity or loading for the climate-control systems (Fig. 7.4). HP-CT X-ray tubes are usually air-cooled because they generate substantial heat during long X-ray exposures. This heat is transferred to the scanning room air, and it needs to be removed efficiently by a cooling system capable of handling the maximum heat produced by the scanner on a hot day in the locale. Typically, manufacturers recommend a room temperature setting of around 22 °C and that this is kept fairly constant throughout the day when the imaging unit is in use as not to affect image quality and equipment function. When the SPECT-CT unit is in standby mode (e.g., overnight), manufacturers usually recommend that the tolerance of variance of about ±4 °C* per hour temperature change is not disregarded as not to damage imaging equipment components (*check with individual manufacturers for specific data**).

For the above reasons, it is usually prudent, when installing a new SPECT/CT system, to ensure that the planning engineer specifies the cooling requirement and tolerances for variation per hour for the scanner to be installed and that the room air-cooling equipment has capacity to deal with this. It might be that a new air-cooling system has to replace an existing one. This is especially important when replacing an existing scanner rather than a complete new build, where the former

Fig. 7.5 Air diffusers (grilles in the ceiling) ensuring uniform air temperature

situation is likely to cause an oversight of the cooling needs, especially within budgetary constraints. It is also important that the conditioned air is ducted correctly throughout the room to ensure that there are no cold or hot areas which might adversely affect the equipment (Fig. 7.5).

Internal cooling may be required for the components located within the CT scanner gantry itself and the associated power distribution unit. Manufacturers of CT equipment may require a chilled water supply, and an external chiller unit should be located somewhere close to the CT suite if required by chosen manufacturer.

7.4 Security

The physical security of nuclear medicine departments housing a SPECT/CT cannot be overstated. There needs to be a robust system for limiting access to controlled areas where radiation may emanate from X- and gamma-rays. The heightened security status of most countries in relation to terrorist activities requires unsealed and sealed sources of radioactive materials to be stored securely and accounted for at all times. Good housekeeping should ensure that the location of each source is recorded on a daily basis together with appropriate records for disposal. Practically, this can be achieved by specifying lead-lined cupboards/storage units when designing a new department and ensuring that these are lockable. It is generally good practice to

have a written procedure for maintaining security within a nuclear medicine department.

7.5 Emergency Call Systems and Life Support

When designing a suite to house a SPECT/CT scanner, it is worth planning where emergency and nurse call buttons should be sited. It is important to ensure that any system for allowing a patient to call a member of staff has enough cable to reach the scanning position on the imaging table (this can be an easy oversight) and that the appropriate alarm signal is transmitted to useful areas where staff congregate to offer prompt assistance; again a shared control room as described earlier has advantages in this respect.

Another aspect of design often neglected when specifying a SPECT/CT facility is the potential requirement to use anaesthetic gases and associated equipment. As techniques develop, the ability of some SPECT/CT systems to perform conventional CT scanning might require facilities that require caring for patients who have sustained trauma, and such patients may be under general anesthesia.

7.5.1 Intravenous Contrast Injectors

Intravenous contrast injectors are discussed in more detail in Chap. 12; however, when designing a SPECT/CT imaging facility, the precise location and likely use of these systems should be planned carefully. They can be floor mounted on wheels or ceiling mounted. Ceiling mounts can offer advantages in terms of improving working space, and they also remove trip hazards, however they can be less flexible should injections need to take place either side of the imaging gantry. Easy mistakes of positioning injectors include:

- Not having enough cable to reach the patient when approaching from either end or side of the scanner gantry.
- Trip hazards for staff – avoid locating near busy thoroughfares.
- Not being in full view of imaging staff when in use when controlling remotely from the scanner control room.

Contrast injectors will be obligatory in most cases where conventional CT imaging is to be performed on a SPECT/CT system and with developing imaging protocols for nuclear imaging in oncology. IV contrast injected via a mechanical pump will be required for some SPECT/CT imaging procedures.

7.5.2 Reporting Station

Experience with hybrid imaging has shown that a team approach to image interpretation is useful. The use of a reporting workstation area within the imaging suite can

allow imaging staff to discuss: cases with clinical colleagues, patient compliance, and potential for and correction of artifacts.

7.5.3 Lighting

Variable-level lighting will be required, provided by a mixture of fluorescent lights and spotlights. Low levels of lighting are required for viewing monitors and laser positioning lights. High levels of lighting are needed for maintenance procedures.

7.5.4 Regulatory Framework

The fundamental task of radiation protection is to minimize or ideally prevent undue exposure of humans and the environment to ionizing radiation. To achieve this end, the International Commission on Radiological Protection [1, 2] has defined the basic principles of radiation protection to be justification, optimization, and limitation of dose. These have been incorporated into the International Basic Safety Standards for Protection Against Ionizing Radiation [3] and form the basis of radiation protection legislation in most countries.

From this, national radiation dose limits should be defined [2] to avoid deterministic radiation damage (injury characterized by a threshold dose and an increase in the severity with dose, e.g., hair loss, cataract) and limit the risk of stochastic effects (the probability of an effect occurring, but not its severity, regarded as a function of dose without threshold, e.g., malignant disease or hereditary effects). These usually consist of annual whole body and extremity dose limits for occupationally exposed persons and members of the public. These can then be used as the basic design criteria for a radiation source installation. For example, in the UK the design of an imaging room may be based on the dose limit for members of the public of 1 mSv per annum. Due to the fact that this could be one of several sources, it has been suggested [4] that the principle of optimization necessitates the adoption of a dose constraint of 30 % of this limit (0.3 mSv per annum). By comparison, the US National Council on Radiation Protection and Measurements [5] suggests a shielding design goal for uncontrolled areas of 0.02 mGy week^{-1} (1 mGy y^{-1}).

7.5.5 Radiation Protection Considerations on Design

The purpose of radiation protection in room design is to protect the department staff, patients undergoing examination, staff working in adjacent areas, visitors, and the general public. The following paragraphs provide a description of the general radiation hazards and controls for CT and nuclear medicine.

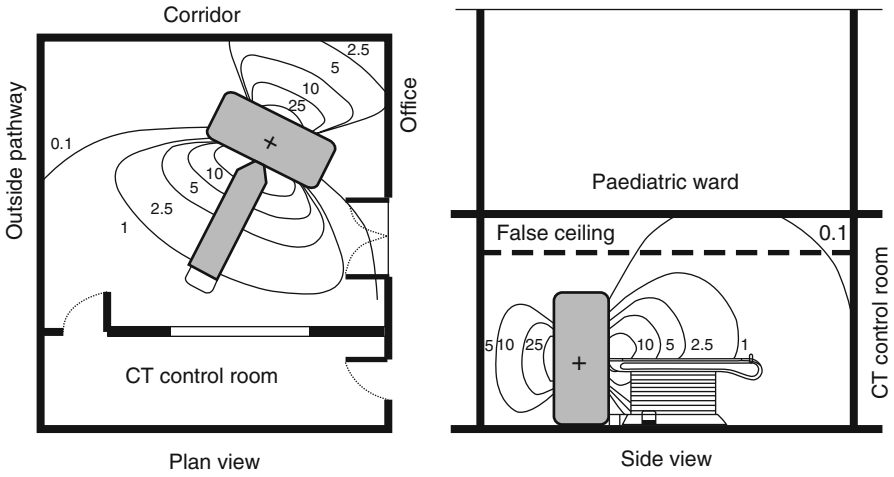

Fig. 7.6 Example radiation dose distribution (μGy per mAs) around a CT scanner in both the plan and side views

7.5.6 CT

As described in detail in Chap. 5, X-ray CT involves the rapid acquisition of image data by rotating an X-ray source in a vertical plane around a patient, who may be simultaneously translated horizontally through the gantry. The primary (or unattenuated) X-ray beam is directed towards the center of rotation, and therefore, only the patient should be subject to the 120–140 kV X-ray beam. Outside of the scanner gantry, the main radiation hazard is due to scattered radiation from the patient and a small amount of leakage from the X-ray tube. Figure 7.6 illustrates the nonuniform nature of the scattered radiation distribution and highlights the need to consider the whole area surrounding a CT scanner, above and below, as well as on the same level. The amount of time these areas are occupied can affect the level of protection required. However, occupancy factors must be applied with caution since the function of adjacent areas can change during the lifetime of the installation.

In order to control the radiation hazard, careful consideration needs to be given to the equipment position and orientation, as well as controlling the access to the area. The level of protection to be provided by the walls, doors, and windows also depends on sufficient information being available regarding equipment workload. For example, it could be envisaged that the SPECT/CT system will be used in diagnostic CT mode for some sessions, leading to a significant increase in scattered radiation levels compared with purely nuclear medicine or hybrid imaging sessions.

ICRP Publication 33 (ICRP 1982) [6] suggests the following general principles for consideration when planning an X-ray room:

(a) That plans should be reviewed by a qualified (radiation) expert
(b) That facilities should be tested after installation for radiation safety

(c) That the maximum workload and number of radiation workers should be taken into account
(d) Access from adjacent areas and designated status of the room and surrounding areas
(e) Placement of protective screens (fixed or mobile)
(f) Orientation of equipment in the room with reference to beam direction
(g) The presence of windows and doors

7.5.7 Nuclear Medicine

Nuclear medicine imaging investigations involve the administration of a radioactive substance to a patient and then subsequent imaging on the gamma camera. The hazards of radioactive substances are twofold: external irradiation from prolonged proximity to the substance and internal irradiation from ingestion or inhalation of the substance through uncontrolled contamination. Facilities should be designed such as to reduce the risks to staff and members of the public from both hazards as much as possible.

Safe storage and handling of radioactive substances and arising waste is crucial in reducing exposure to staff and members of the public. To this end, it is essential to have robust working procedures, good training of relevant staff, and well-designed facilities. The latter aspect is the one that is relevant to this chapter.

Radioactive substances for nuclear medicine imaging, such as Mo99/Tc99m generators, are normally delivered to the hospital by commercial suppliers. Delivery may be to a radiopharmacy or directly to a nuclear medicine department, depending on the available resources and relevant licenses. In either case, suitably secure and shielded storage areas are essential to minimize doses. For example, in the case of a Mo99/Tc99m generator, installation directly into a shielded housing within an appropriate isolator is ideal. These are constructed with adequate shielding and allow ease of access to the generator for elution; the eluate itself is then available in the isolator to be used for preparing the relevant radiopharmaceuticals. Adequate storage of other delivered radionuclides will depend on their activity and the gamma-ray energy of the radionuclide. High-energy radionuclides will need substantially more shielding. Lead-lined cupboards are available, but sub-containment within lead pots may be necessary to provide the additional shielding required for high-energy gamma emitters. These cupboards should be situated on the ground, due to their weight, and should be sited in designated radiation areas away from highly occupied locations. Their position, however, should be convenient enough to avoid staff having to carry radioactive substances over significant distances, which could otherwise cause unnecessary radiation exposure and increase the risk of dropping the substance and causing contamination.

Security of radioactive substances is also of paramount importance in order to avoid theft and consequent misuse. Depending on the activity and radionuclides involved, there may be a need to have at least two levels of physical security in place for storage facilities for radioactive substances. Access to keys or codes should be restricted as far as possible, and keys should be stored in a secure area away from the storage facility.

The layout of rooms within the SPECT/CT unit should be such as to allow a logical flow of radioactive substances and patients to minimize transfer and its associated radiation protection risks. The unit should be reasonably self-contained so that there are not large distances over which radioactive substances have to be transferred.

The waiting area for patients who have been administered radiopharmaceuticals should ideally be sited away from sensitive areas, for example, pediatric departments and obstetric ultrasound. There should be adequate shielding in the walls of the waiting area, depending on its position and the occupancy of the surrounding areas, to ensure that doses to persons in these places are below relevant dose constraints.

There should also be sufficient shielding in the perimeter walls of the SPECT/CT scanning room (attenuating gamma rays) to ensure that doses to persons outside of the unit are kept below appropriate dose constraints. Assessment of shielding requirements will need to take into account workload, radionuclides used, and radioactivity handled and also occupancy of areas outside of the unit. The shielding around the perimeter of the scanning room required for the CT component will normally be sufficient to attenuate the external irradiation from radiopharmaceuticals and radioactive patients.

There should be a designated toilet for nuclear medicine patients close to the waiting area and ideally within the confines of the unit so as to minimize the risk of spread of contamination into nondesignated areas. The toilet should be clearly labeled as being for nuclear medicine patients only. Drainage from this toilet, and from any sinks into which radioactive substances may be discharged, will need to be carefully considered. Drainage should be as direct as possible into the hospital sewerage system. Depending on the radioactivity to be discharged, it may be necessary to advise local estates staff and to label potentially radioactive drainage routes so that appropriate actions can be taken in the event of blockages. It will not normally be necessary to install holding pits, designed to delay release of radioactive aqueous waste, for diagnostic radionuclides.

Radiopharmaceuticals should be stored in the injection area for ease of access. There should also be separate shielded secure facilities for the storage of radioactive waste close to the point of production. All storage facilities should be clearly labeled as containing radioactive substances.

All floors, walls, and work surfaces in the SPECT/CT unit need to be easy to decontaminate in the event of a spillage.

Sinks should be provided to allow disposal of radioactive substances and, separately, for handwashing. The former should be clearly labeled as a designated sink for disposal and the latter labeled as not being for disposal. Sinks for handwashing should have elbow- or knee-operated taps or have electronic sensor operating so that contaminated hands can be washed without causing contamination of the taps. Sinks designated for disposal should have the means to be able to be left running for prolonged periods so that radioactive substances are properly flushed into the drainage system.

Workbenches should be of a sufficient size as to allow for uncluttered working to reduce risk of contamination and to simplify decontamination if necessary.

Equipment such as lead bench guards should be provided to minimize external irradiation during preparation of patient doses. A radionuclide calibrator for radioassay of patient doses prior to administration should be in close proximity to the preparation workbench, as should a shielded radioactive waste container to avoid unnecessary transfer of radioactive substances.

If lung ventilation scanning is to be performed with radioactive aerosols, it will be advantageous to have ceiling-mounted extraction fans to reduce contamination of equipment and personnel.

All staff must monitor their hands for contamination before leaving designated radiation areas, and a wall-mounted contamination monitor is the ideal way of providing this control.

All designated radiation areas must be clearly labeled to identify the potential radioactive substance hazard.

7.6 Room Design Specifics: Recap

7.6.1 CT Aspects

- During the procurement process, the equipment supplier should be instructed to provide X-ray radiation scatter diagrams, including the scanning conditions used to acquire the data (e.g., kV, slice thickness, number of contiguous slices). It is also essential at the outset to determine the usage of the equipment since if it is to be used for diagnostic CT sessions, this requires additional planning and protection.
- The walls (internal and external), floor, ceiling, windows, and doors (including frames) will require some level of protection from the X-ray radiation, for example, lead lining/specialist plaster/leaded plywood. Protected doors should open into the scanning room, providing shielding for people who accidentally enter the area during scanning with X-radiation.
- All entries to the scanner room require suitable warning indicators, such as illuminated signs to indicate when X-ray radiation is being produced, and fixed signs outlining the hazard and limitations on access.
- Personal protective equipment, in the form of lead (Pb) or lead-equivalent aprons, should be provided if staff or comforters or carers are required to remain in the scanner room during an X-ray examination. Suitable hangers may be required to prevent creases forming or other damage occurring due to folding of the aprons.
- Due to the levels of scattered radiation for CT acquisitions, staff generally operate the equipment from a separate control room. The control room should be designed to allow the operator to observe the patient throughout the CT acquisition, preferably by direct viewing, but high-resolution CCTV cameras mounted to display both sides of the gantry aperture may suffice. They should also be able to see the entrances to the room to prevent unprotected personnel entering the

room during exposures. Users often prefer a separate entrance from the control room to the scanner room rather than using the main entrance.
- Access to the CT scanner room should be controlled and/or authorized from the control room; the door to the control area from the main staff or public space should be lockable.

7.6.2 Nuclear Medicine Aspects

- There is a potential for radioactive contamination in the scanning room; some patients will require injection of the radiopharmaceutical in this room, for example, in the case of dynamic acquisitions, and contamination may also arise from urine or blood spills. Therefore, the finishes of floors and work surfaces must be such as to allow ease of decontamination; that is, they should be smooth and impervious. Floors should be made of hard-wearing vinyl, with close attention to joins (these should be heat sealed), and should be coved up the walls. Walls should be painted with nonabsorbent paint. Worktops should be constructed of materials such a melamine which need to be well sealed at joins with sinks and walls. At least one sink should be available for personnel decontamination in the event of a spillage. All gaps between the worktop and adjoining surfaces and sinks should be well sealed to prevent radioactive substances becoming lodged.

7.6.3 Injection Room

- This is the main area in which radiopharmaceuticals are handled, and so there is a potential for both contamination and external irradiation.
- As with the scanning room, floors and work surfaces need to be designed in order to minimize the likelihood of contamination arising and the ease with which they can be decontaminated in the event that contamination does occur. It is preferable to have two sinks available in the injection room, one for personnel decontamination and the other for disposal of aqueous radioactive waste. The drainage routes for these sinks need to be considered; that is, it is preferable that pipes run as directly as possible into the ground in order to reduce external exposure from radioactivity in above-ground pipes.
- The avoidance of clutter is essential in maintaining control of contamination, and this is best achieved through careful consideration of storage facilities. For example, wall- and floor-mounted cupboards are ideal for storage of syringes, needles, cannulae, and water for injection.
- To minimize external irradiation of staff handling radioactive materials, the room should be equipped with adequately shielded bench guards for preparing patient doses, and shielded cupboards for radiopharmaceutical and radioactive waste storage. Consideration must be given to shielding the walls of the room itself to

minimize exposure to people in adjacent areas and cross talk with gamma cameras nearby. The amount of shielding required will depend on activities handled, local shielding, and the type and occupancy of adjacent areas.

7.6.4 Waiting Areas

- There is a potential for contamination from, for example, urine or blood spills so materials for floors, chairs, and tables should be carefully chosen so as to aid decontamination if necessary. Carpets, for example, should be avoided, and chairs should be nonabsorbent vinyl.

7.6.5 Patient Toilets

- Toilet facilities used by nuclear medicine patients are very likely to become contaminated since urinary excretion is the main route of elimination from the body. It is essential, therefore, that flooring is well-sealed, hard-wearing vinyl in order to ease decontamination when necessary. Similarly, the toilet and sink need to be well maintained and have good seals with the floor and walls.
- Clear signage is important to prevent the use of these toilet facilities by nonnuclear medicine patients. Signs may be required from time to time to restrict access completely in the event of a significant spill, where the risk of spreading contamination is high, until such time that the radioactivity has decayed to safe levels.

There is always likely to be variable advice and interpretation of international and national radiation protection schemes; however, the general advice and information given above should assist imaging personnel to consider key areas when designing a section to house a SPECT/CT scanner, in consultation with relevant architectural and radiation protection professionals.

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Part III
Clinical SPECT/CT

Chapter 8
Clinical Utility (Applications) of SPECT/CT

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8.1 SPECT/CT in Oncology, Endocrine Imaging, and Infection

Radionuclide imaging has for many years had an established role in oncological imaging. This includes the conventional bone scan using diphosphonates, as well as a range of techniques using tumor-specific radiopharmaceuticals. In this chapter, some of these agents will be described in more detail. As in other branches of radionuclide imaging, the clinical utility of these types of radiopharmaceutical tracers has been improved by the increased resolution of gamma camera systems and subsequently by the development of single photon emission tomography (SPECT). The increasing use of SPECT to produce tomographic images that could be correlated with other cross-sectional imaging techniques has made tumor-specific imaging much more clinically acceptable and in this has paralleled the increasing use of FDG PET in clinical practice.

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The emergence at the beginning of the twenty-first century of commercially available hybrid PET/CT devices was the result of both technical and clinical demands for more rapid and accurate attenuation correction but led to the rapid introduction into clinical practice of hybrid fusion imaging demonstrating both physiological metabolic activity by means of FDG uptake and also anatomical and morphological information as CT slices. Within a very short period, a large body of evidence was developed showing that the hybrid display of both PET and CT data on a single image set improved both diagnostic accuracy and reporting confidence and helped establish FDG PET/CT as a mainstream imaging tool in cancer management. In parallel with this came the development of hybrid SPECT/CT gamma camera systems and the ability to produce identical hybrid image displays of "conventional" radionuclide tracer distribution. The same increase in diagnostic accuracy and reporting confidence would therefore be expected, although there have been few if any publications specifically investigating this issue relating to non-PET/CT tracers. An overview of the available evidence was published by the International Atomic Energy Authority in August 2008 [1], which concluded that the impact of hybrid SPECT/CT remained subjective, as the concept of incremental confidence was difficult to quantify; however, the likelihood that hybrid imaging would produce greater credibility was accepted.

In this IAEA document, the clinical indications for use of SPECT/CT in preference to conventional radionuclide imaging were stated to be:

- High suspicion for active disease, where multiple sites may be localized to aid definition of extent of disease.
- Planning of treatment.
- Monitoring of response to treatment.
- Assessment of abnormal structural findings on anatomical cross-sectional imaging which are of equivocal functional significance.
- Assessment of patients with high clinical suspicion but with absence of structural pathology on conventional imaging. This assessment might follow inconclusive results not only from anatomical imaging such as CT or MR but also inconclusive results from other planar radionuclide imaging.

It can therefore be argued that there is a clinical indication for the use of SPECT/CT in any radionuclide tracer study in oncology, where not only is there a requirement to detect abnormal tracer activity but a need to anatomically localize this, either to confirm the pathological nature of the uptake or to target therapy. This latter requirement accepts that, for virtually all gamma camera systems, the spatial resolution that can be achieved is in the order of several millimeters for planar imaging and, for SPECT, this is only marginally improved, whereas CT images have a much greater spatial resolution and allow image displays showing normal anatomical landmarks and surrounding structures. This much greater spatial resolution is achieved not only on high-specification SPECT/CT devices with full multislice helical CT gantries but also with the low-specification attenuation correction SPECT/CT systems as typified by the GE Hawkeye.

8.2 Cancer

8.2.1 ^{111}In: Octreotide Imaging of Neuroendocrine Tumors

Imaging of known or suspected neuroendocrine tumors with octreotide is probably the most widely used tumor-specific agent in the UK. The neuropeptide somatostatin was first identified in 1973 and has since been recognized as a major neurotransmitter and neuromodulator (with inhibitory regulation of exocrine secretions, smooth muscle contractility, and absorption of nutrients).

Human somatostatin appears in several organs including the hypothalamus, pituitary gland, GI tract, pancreas, and immune system. The action of somatostatin is mediated via five high-affinity somatostatin receptors (SSTRs) located on the plasma membrane of target cells. Five human subtypes have been cloned and partially characterized. All five receptors bind human somatostatin-14 with high affinity, but the biological short half-life of somatostatin in vivo of only 2 min prevents its application in clinical practice.

To overcome this, synthetic analogs of somatostatin-14 consisting of 8 amino acids were developed with a biological half-life of several hours, and of these, octreotide is the one most clinically relevant. Octreotide is bound with high affinity by SSTR2 and SSTR5 receptors and modestly by SSTR3 but not bound at all by SSTR1 and SSTR4. GI tract neuroendocrine tumor cells belong to the group of disseminated neuroendocrine cells. These are cells that are scattered throughout the mucosa of the gut and form the islets in the pancreas. These cells have many histologic similarities to cells of neural origin. They constitute the so-called diffuse endocrine system. Misunderstanding was perpetuated by pathologists who coined the term carcinoid tumors in 1963; the terms neuroendocrine tumors and neuroendocrine carcinomas were chosen in the end for common usage in the new WHO classification of 2000 to avoid confusion, but the term carcinoid is not completely abandoned as this is defined as a well-differentiated NET tumor.

GI-NE tumors are the most frequent that express SSTRs with high density. Others include medullary thyroid cancer, small-cell lung cancer, meningioma, medulloblastoma, glioma, breast cancer, lymphoma, and renal cell cancer. Pancreatic NETs (e.g., gastrinomas, glucagonomas, and VIPomas) and gut NETs usually express SSTRs in 80–100 % of the cases. Insulinomas, however, have a lower incidence of SSTR expression (about 50 %) and are therefore more difficult to detect on SSTR imaging. SSTR subtype 2 is the most frequently expressed in GEP-NETs, while SSTR4 is rarely detected. Undifferentiated NETs express SSTRs less often.

It is currently difficult to visualize NETs completely on standard imaging techniques such as ultrasound, CT, and MRI because such tumors are frequently small and disseminated over several organ systems and are therefore beyond the craniocaudal limits of scanning.

Imaging may be needed to detect primary neuroendocrine tumors, where patients present with clinical or biochemical signs of carcinoid syndrome, or, more commonly, to assess possible tumor spread following a surgical diagnosis. However,

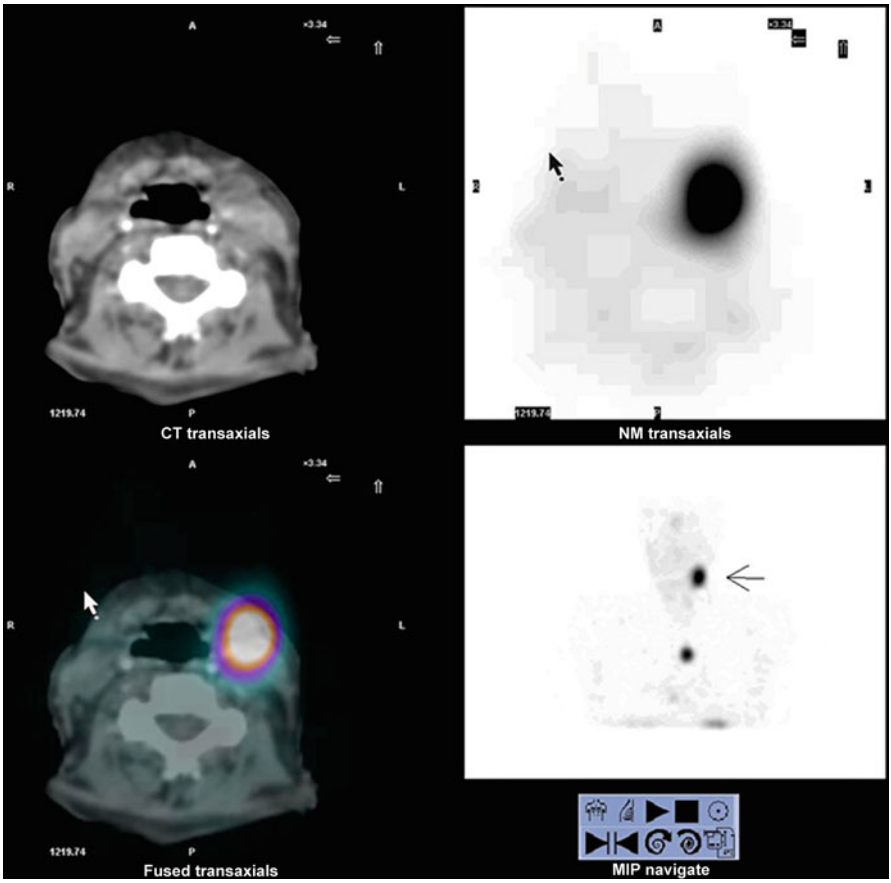


Fig. 8.1 Octreotide scan. Patient presenting with labile hypertension and palpable mass in left side of neck. CT had shown bilateral vascular masses at the carotid bifurcations; the larger right-sided lesion had been excised confirming a primary carotid body tumor. Prior to second surgery, scanning was undertaken to assess the left-sided mass. *Image 1* shows intense uptake in the left carotid body tumor (*arrow* on MIP image) with another focus seen on MIP. (*Top right*) SPECT image; (*bottom right*) MIP image; (*top left*) low-dose CT (*liver* windows); (*bottom left*) SPECT/CT

octreotide has been shown to have a suppressant effect in such tumors, and diagnostic scanning may be requested to assess suitability of known tumors for such therapy and subsequent to therapy to assess response [2].

Although highly sensitive, with reported sensitivity of 82–95 %, the specificity of octreotide imaging is limited firstly by the low spatial resolution of conventional nuclear medicine imaging and also by the normal distribution of octreotide within the body. This includes variable bowel excretion and physiological liver and spleen uptake; octreotide uptake is also reported in benign disease such as sarcoidosis [3]. Further difficulty is encountered by the variability of primary tumor location within bowel. For these reasons, accurate anatomical localization is needed to confirm both nature and location of focal tracer uptake. A number of studies [4–6] have shown that interpretation of SPECT/CT images improved diagnostic certainty in octreotide

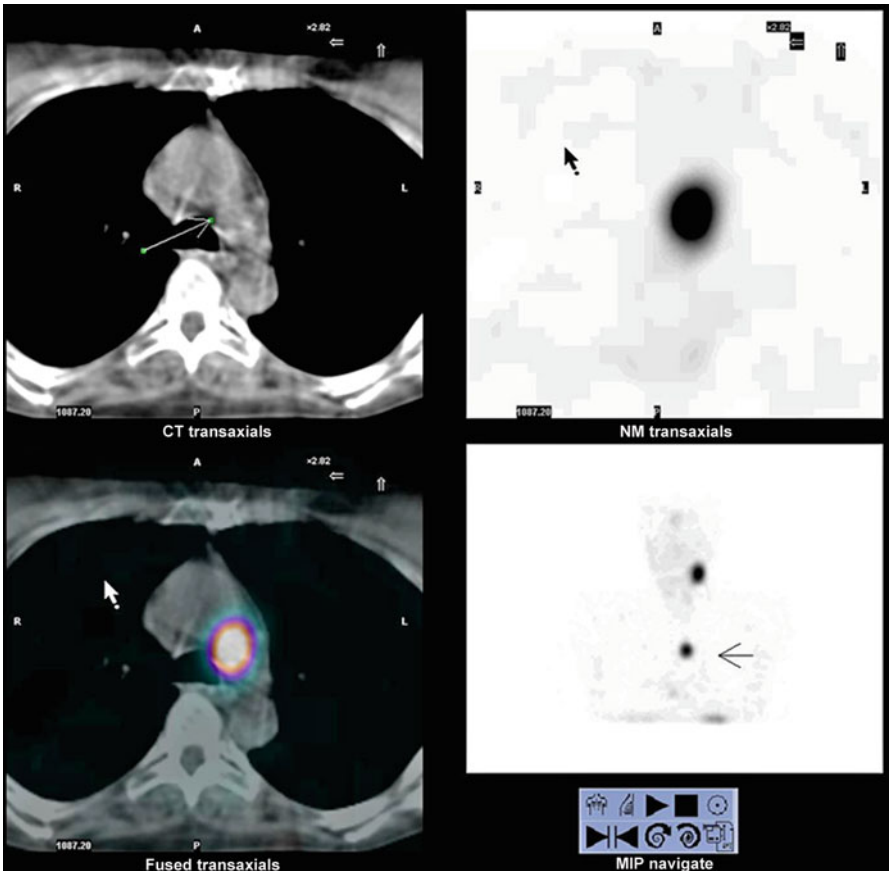


Fig. 8.2 Octreotide scan. *Image 2* shows a second intense focus of activity (*arrow* on MIP image) associated with a small nodule in the aortopulmonary window (*arrow* on CT image). Review of the previous CT revealed a highly vascular mass. (*Top right*) SPECT image; (*bottom right*) MIP image; (*top left*) low-dose CT (*soft tissue* windows); (*bottom left*) SPECT/CT

scanning from 57–61 to 91 % and localization of liver metastases from 45–58 % certainty to 100 %. This has led to an overall improvement in clinical effectiveness of this technique, and in one study [6], SPECT/CT was considered to outperform both SPECT and conventional CT in diagnostic accuracy. Figures 8.1 and 8.2 illustrate accuracy of anatomical localization using SPECT/CT in octreotide imaging.

Octreotide imaging does have some limitations, largely due to the biological variability of neuroendocrine tumors and the variation in somatostatin receptor expression; not all tumors are therefore detected by this tracer, but the routine use of SPECT/CT in indium octreotide scanning is advised particularly for imaging of the bowel, retroperitoneal structures, and liver.

Other neuroendocrine tracers exist; indium-labeled DTPA pentetreotide has similar properties to octreotide but is not currently used in mainstream clinical practice,

and a specific somatostatin 5 receptor binding agent, depreotide, was developed which can be labeled with technetium. Depreotide was briefly commercially available for the imaging of non-small cell lung carcinomas using SPECT. Clearly the arguments for improved anatomical localization would also apply for these agents.

8.3 MIBG Imaging of Paragangliomas

^{123}I metaiodobenzylguanidine (MIBG) has been used in clinical practice for many years to localize paragangliomas; the two major indications have been the detection and localization of pheochromocytoma in adults and the staging of neuroblastoma in pediatric oncology. MIBG is a precursor chemical for norepinephrine, which is synthesized in normal adrenal tissue and over expressed in certain neural crest tumors as above. Preoperative localization of pheochromocytoma by MIBG scanning is well established; while these tumors are most commonly seen in the adrenal gland, they can occur in ectopic locations both in the chest and abdomen and in rare cases may be multiple, particularly when these tumors present in a multiple endocrine neoplasia syndrome or as part of the von Hippel–Lindau spectrum of disorders. Accurate anatomical localization of an ectopic pheochromocytoma would be expected to improve surgical management; however, a recent paper [7] showed planar MIBG imaging to have a specificity of 82 % and only minimal improvement in this by use of SPECT, though reader confidence was increased. Figure 8.3 shows an example of MIBG scanning in pheochromocytoma.

In pediatric practice, neuroblastoma staging involves a combination of MR scanning and nuclear medicine imaging with both MIBG and MDP; conventional CT and PET/CT have both been used, but there is little published evidence on the use of SPECT/CT; the increased radiation dose may possibly be a factor in this.

There are also case reports of MIBG use in imaging of adult paragangliomas; Figs. 8.4, 8.5, and 8.6 show an example of this.

8.4 ^{131}I Imaging of Thyroid Cancers

Although relatively uncommon, with an incidence of approximately 1 in 10,000, well-differentiated thyroid carcinoma is a disease of younger patients and is being increasingly recognized and treated. Standard therapy includes total thyroidectomy and postsurgical ablation of residual disease with ^{131}I-iodide. Prognosis depends on the stage of disease at presentation, being worse in the presence of nodal or distant metastasis, and is also worse in dedifferentiated tumors as these show reduced trapping of iodine. Current practice is therefore to perform a diagnostic iodine scan prior to iodide therapy to assess whether tumor deposits trap iodine. Accurate anatomical localization is not usually required for this. However, because of the prognostic significance,

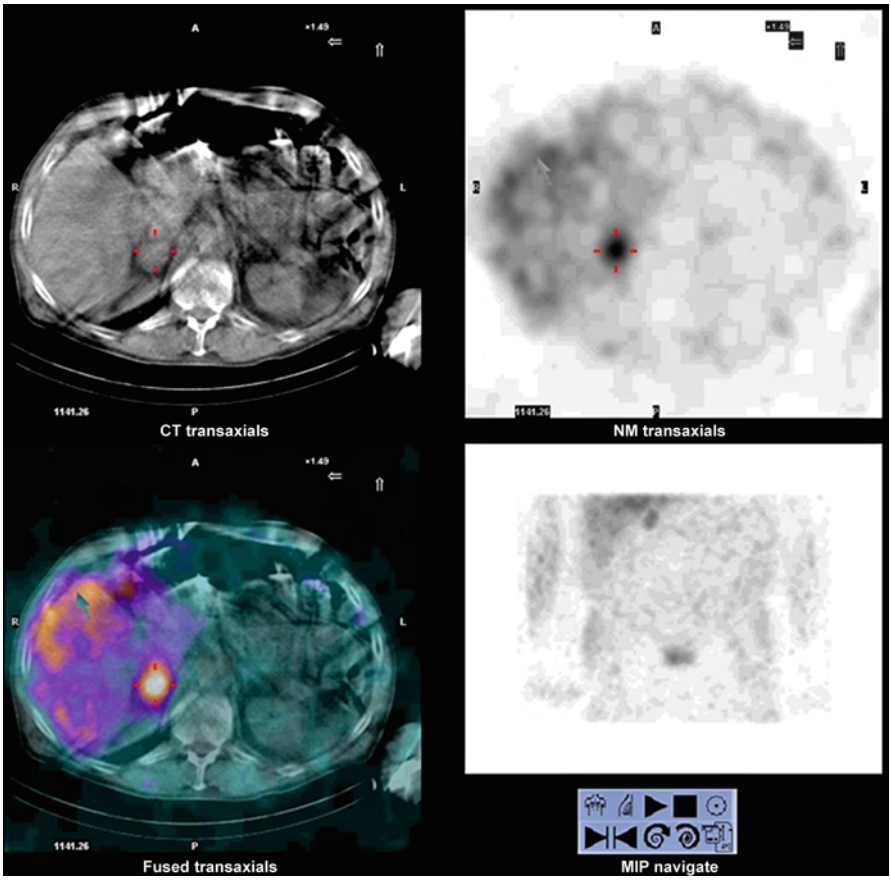


Fig. 8.3 MIBG scan. Patient with resistant hypertension and elevated urinary catecholamines, suggesting a pheochromocytoma. Conventional CT had shown a small right adrenal mass. MIBG SPECT/CT confirms intense uptake in this adrenal mass (crosshairs) typical of pheochromocytoma, confirmed at subsequent surgery. (*Top right*) SPECT image; (*bottom right*) MIP frontal view; (*top left*) low-dose CT (bone windows); (*bottom left*) SPECT/CT

accurate nodal staging may become increasingly important, particularly as this may influence the inclusion of nodal dissection in the primary operation [8]. Two recent studies [9, 10] assessed the impact of SPECT/CT iodine imaging at initial assessment; one concluded that the increased sensitivity of SPECT/CT delivered a high negative predictive value for nodal recurrence, while in the earlier paper, a change in staging was seen in 25 % of patients with a resultant change in treatment. Equally, assessment of possible residual disease following initial therapy will impact on the nature and frequency of follow-up investigations. Another recent publication [11] assessed the impact of posttreatment 131I SPECT/CT on management and showed a change in

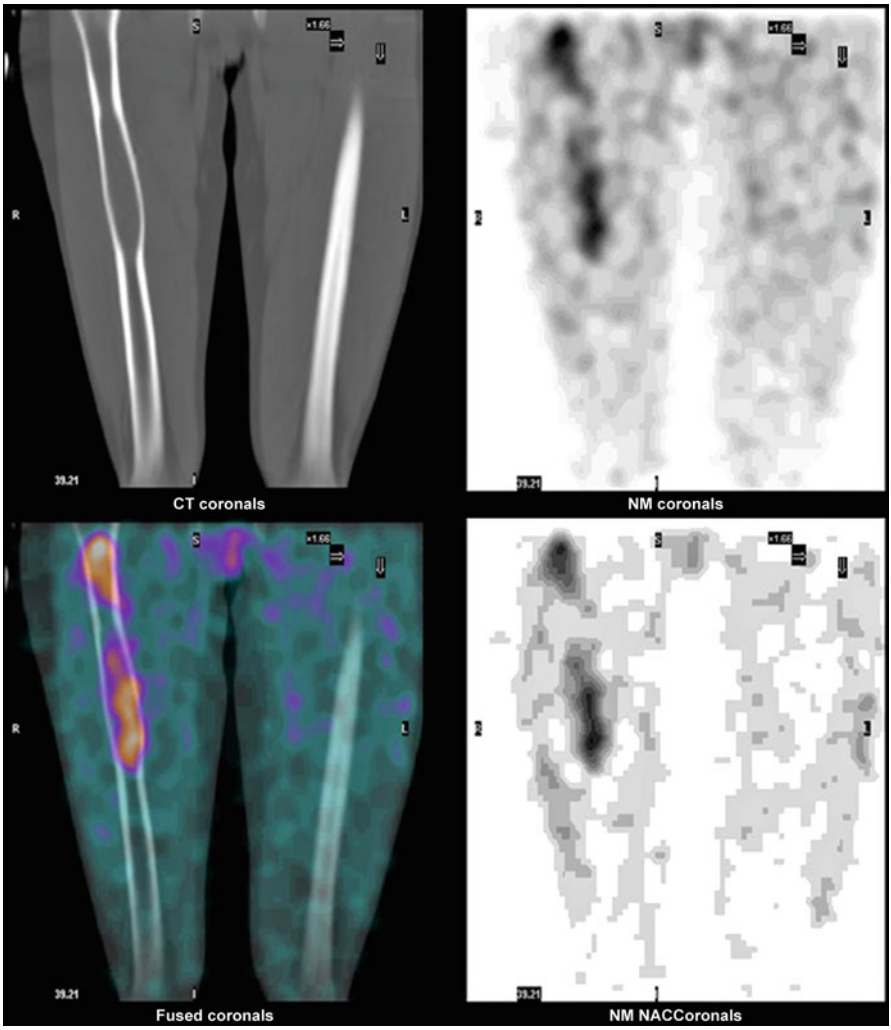


Fig. 8.4 MIBG scan: Patient with previous history of excision of malignant paraganglioma. Presents with pain in right thigh; X-rays (not shown here) demonstrate lytic expansile lesion in right femur. Urinary catecholamines suggest tumor recurrence. Scan requested to assess whether lesion contains functional tumor and if further lesions are present. *Image 1* shows MIBG uptake in the marrow cavity of the right femur; CT component demonstrates the lytic expansile mass, and fusion SPECT/CT confirms the uptake to be in this expansile marrow cavity lesion. (*Top right*) SPECT image; (*bottom right*) planar image; (*top left*) low-dose CT (bone windows); (*bottom left*) SPECT/CT

postoperative staging in 6 % of patients, but in 20 %, the accuracy of staging provided by SPECT/CT reduced the need for alternative imaging.

Following therapy, patients are monitored by thyroglobulin assessment, this being an extremely sensitive biochemical marker for thyroid cancer. However, detection of a rising thyroglobulin level requires confirmation of the site and extent of recurrence and again may require assessment of iodide trapping if repeat ablation

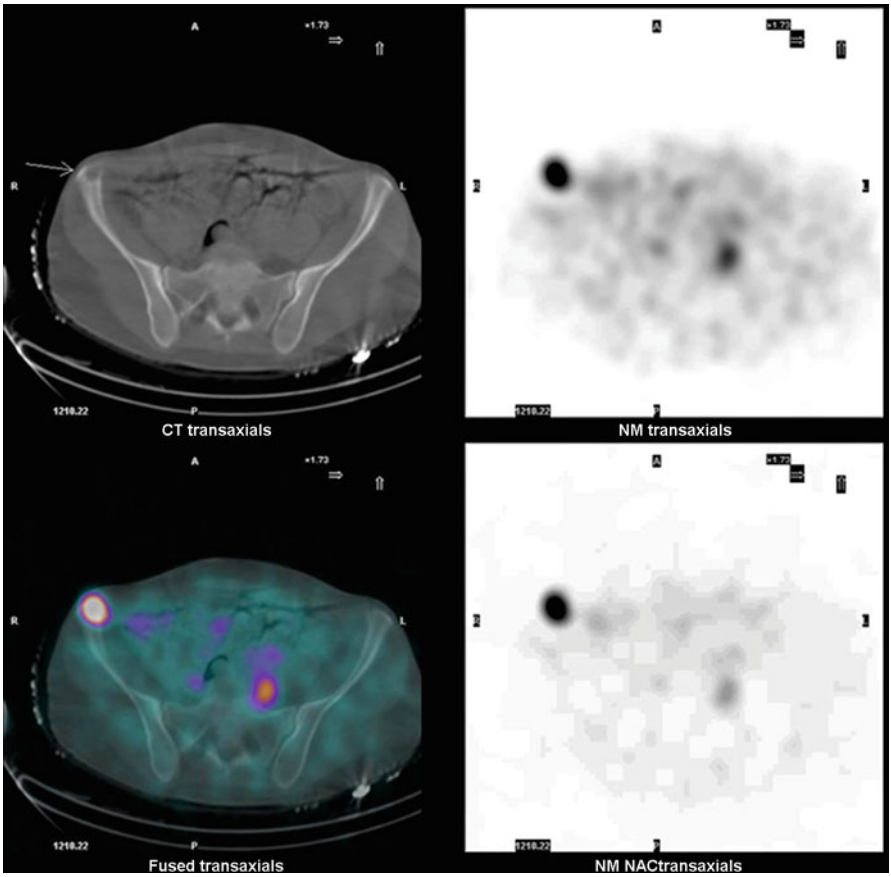


Fig. 8.5 MIBG scan. *Image 2* shows an unexpected highly MIBG avid lesion in the right anterior superior iliac crest; SPECT/CT confirms uptake to be in a small lytic bone lesion (*arrow*). (*Top right*) SPECT image; (*bottom right*) SPECT without attenuation correction; (*top left*) low-dose CT (bone windows); (*bottom left*) SPECT/CT

therapy may be offered. This assessment is usually made with radioiodine scanning, after withdrawal of thyroxine replacement therapy, although there is a move toward using FDG PET/CT as an alternative. As with other tumor localization techniques, the increased anatomical accuracy of SPECT/CT can significantly improve the accuracy of interpretation, in one recent study [12] by up to 67 % of patients.

8.5 Prostate Cancer

Although one of the more commonly seen malignancies, the staging and follow-up imaging of prostate cancer heavily depends on MR scanning of the primary tumor and the use of conventional bone scanning with HDP or MDP for assessment of

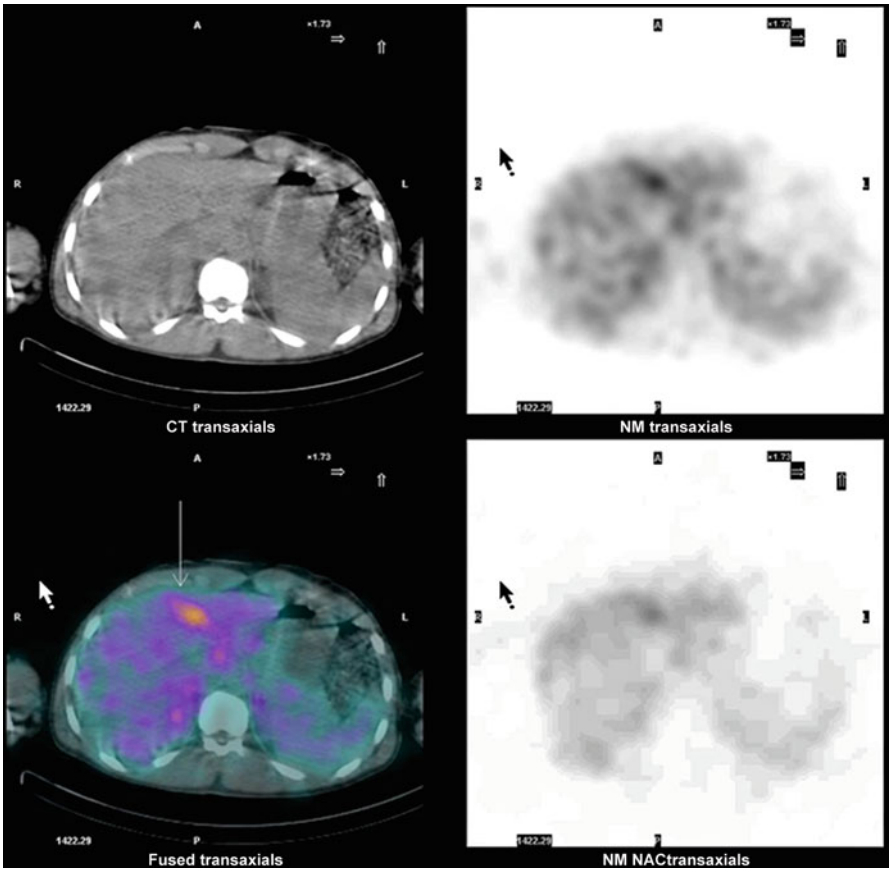


Fig. 8.6 MIBG scan. *Image 3* shows a small focus of MIBG uptake in segment 4a of the liver (*arrow*); suspicious of a further metastatic deposit; no CT abnormality is shown. (*Top right*) SPECT image; (*bottom right*) SPECT without attenuation correction; (*top left*) low-dose CT (bone windows); (*bottom left*) SPECT/CT

skeletal metastasis. Diagnostic difficulties do however arise, particularly in the assessment of equivocal lymph node enlargement detected on either MR or conventional CT scanning, as well as in the assessment of PSA rise occurring posttherapy. A number of tracers have been investigated for the assessment of patients in these groups, and currently two appear to show some promise: ^{11}C- or ^{18}F-labeled choline, as a PET/CT tracer, and ^{111}indium-labeled capromab pendetide. This latter has been reported to be significantly more sensitive than CT or MR in detection of lymph node metastasis [13], and a further improvement in false-positive rates by 46 % using SPECT/CT has been claimed [14].

There has also been research work in the use of this tracer (as there has been for choline PET) in pretreatment localization of primary tumor within the prostate prior to intensity-modulated radiotherapy (IMRT) [15] and recently in assessing the grading (aggressiveness) of primary prostatic tumors [16].

8.6 Sentinel Node Imaging

The concept of the sentinel lymph node has become firmly established in the surgical management of a number of tumors. This concept is that if the first draining node (sentinel node) of a primary malignancy can be identified, and then surgically sampled, the absence of histological involvement of that node allows the surgeon not to undertake further nodal clearance, while a positive histological diagnosis will lead to more extensive nodal clearance and/or further adjuvant treatment. In the UK, the ALMANAC study established sentinel nodal sampling as standard practice in the management of breast carcinoma, and most centers now practice sentinel node scintigraphy either with planar gamma camera imaging or with intraoperative gamma probes. However, sentinel node scintigraphy is becoming established practice in a range of other tumors; currently this includes malignant melanoma [17], and there is increasing interest in its use in pelvic malignancies including penile, cervical, and vulval carcinoma [18], and prostatic carcinoma, as well as head and neck tumors.

In breast carcinoma, the standard technique involves the subdermal injection of sulfur colloid and dynamic imaging with a planar gamma camera, with subsequent probe localization and skin marking of the sentinel node. For most patients, this will lie in the axillary drainage, and subsequent surgical localization is not considered difficult. However, in a proportion of patients, the sentinel node may lie in a deep location, making this difficult to localize, or in an unexpected location. This has led to the use of SPECT/CT to assess equivocal scans, and there have been a number of papers reporting significant improvement in preoperative localization of both deep axillary and non-axillary nodes, to levels approaching 100 % [19, 20].

While in breast carcinoma SPECT/CT may be seen as a problem solving tool, in other tumors the variable location and potential difficulty in surgical localization of sentinel nodes have led to the advocation of routine SPECT/CT [21]. In melanoma, the prognosis is heavily dependent on the size and invasiveness of the primary lesion and the presence at diagnosis of detectable spread. Thus, detection of the sentinel node can be of significant importance. However, the very variable lymphatic drainage of skin lesions makes localization difficult. The routine use of SPECT/CT has significantly increased the accuracy of preoperative nodal localization (Herbert JC, personal communication) [22] and thus impacted on the clinical management.

Similar considerations apply to pelvic tumors. Sentinel node imaging is becoming established in the management of patients being considered for radical surgery both for penile carcinomas in males and for cervical and vulval carcinomas in females. In these pelvic tumors, the sentinel node may be located along the iliac chain or in the retroperitoneum, and successful localization can be extremely difficult without the additional anatomical information available from SPECT/CT. Accurate localization however facilitates surgical localization and potentially allows laparoscopic minimally invasive sampling [23]. Interest is now developing in this technique for operative staging of prostate cancer, where surgical lymphadenectomy is advocated by some surgeons as part of radical prostatectomy and nodal staging may influence the use of IMRT [24].

Sentinel node imaging also shows promise in the management of head and neck cancers. Currently most centers will undertake radical neck dissection as part of the surgical management of oropharyngeal tumors, but reported studies suggest this yields positive nodal spread in less than 50 % of patients; this has led to the suggestion that sentinel node imaging and selective dissection would carry much lower postoperative morbidity. The use of SPECT/CT to accurately define the sentinel node is considered to significantly improve this process [25].

8.7 Endocrinology

8.7.1 Parathyroid Adenoma Localization

In recent years, there has been a significant increase in the demand for parathyroid imaging. This has been due to a number of factors. Firstly, the increased use of biochemical multichannel analyzers, introduced into most hospitals in the 1990s, has led to a dramatic increase in the number of patients being diagnosed with hypercalcemia and simplified the biochemical diagnosis of primary hyperparathyroidism – that is, patients with abnormal calcium metabolism, raised parathormone levels, and normal renal function. Secondly, there has been an increasing clinical awareness among endocrine physicians that surgical treatment of primary hyperparathyroidism can produce significant improvement in well-being and reduced morbidity in the group of patients previously regarded as having "subclinical" disease. Thirdly, in renal medicine, there has been a parallel recognition of the potential benefit of treating patients with renal disease who develop parathyroid disease secondary to the alteration in calcium metabolism caused by renal failure, particularly those who develop autonomous parathyroid gland hyperplasia – so-called tertiary hyperparathyroidism.

The final factor has been a shift in clinical surgical practice: Traditionally the surgical management of hyperparathyroidism had been to explore both sides of the neck, exposing all four parathyroid glands and removing any that appeared pathologically enlarged. Because the normal parathyroid gland is extremely small and biochemically significant adenomas can weigh less than 1 g, there was a recognized incidence of

failed exploration and reoperation following radiological localization. However, over the last 10 years, there has been a move firstly to unilateral neck exploration and more recently to single-gland exploration, often through an incision of less than 2 cm. This change has been driven by the significantly lower surgical morbidity associated with minimally invasive surgery and the much shorter postoperative hospital care needed; in some centers, this is now being looked at as day case surgery.

To facilitate this surgical change, accurate preoperative localization is required [26, 27]. As noted above, in the normal individual, there are four parathyroid glands, usually lying alongside the thyroid and supplied by branches of the middle thyroid artery (superior parathyroid) and inferior thyroid artery (inferior parathyroid), each weighing less than 50 mg. Adenoma can arise in any one gland and also in a small but important proportion of cases in an ectopic location anywhere between the skull base and the aortic arch. Nuclear medicine localization techniques have existed since the 1970s, but the older technique of dual isotope subtraction imaging, using either thallous chloride and ^{99m}Tc pertechnetate or thallium and radioiodine, has now largely been replaced by dual time point imaging with ^{99m}Tc methoxy-isobutyl isonitrile (MIBI), with image acquisition at 15 and 120 min. This technique was shown to have a greater sensitivity and moderately better specificity; greater overall accuracy however was claimed by the routine use of SPECT [28]. Despite this, many surgeons required more accurate anatomical localization for single-gland surgery, which in most centers is achieved by targeted preoperative neck ultrasound following nuclear medicine imaging. Ultrasound however cannot accurately localize ectopic tumors, particularly in the mediastinum.

The potential for SPECT/CT with fusion imaging to accurately localize ectopic parathyroid tumors was recognized relatively early [29], but there are now several papers advocating SPECT/CT acquisition routinely as one of the two time point acquisitions, with the intention to provide a single image and reduce the need for associated ultrasound scanning [30–32]. This has not yet been proven to be clinically effective in terms of surgical outcome [29], but further work on this is advised [31] and may yet provide such proof.

Figures 8.7 and 8.8 show examples of SPECT/CT localization of parathyroid adenoma; Figs. 8.9 and 8.10 show how SPECT/CT can localize ectopically positioned glands in the neck, and Figs. 8.11 and 8.12 illustrate a mediastinal ectopic adenoma.

8.8 Imaging of Infection

Nuclear medicine techniques have been used for many years to image infection and inflammatory diseases, with varying success. A number of tracers have been utilized, of which the longest established has been ^{67}gallium citrate, which is still used occasionally and may have a role in the imaging of distal limb infection in diabetes. However, in recent years, nuclear medicine imaging of suspected infection has been

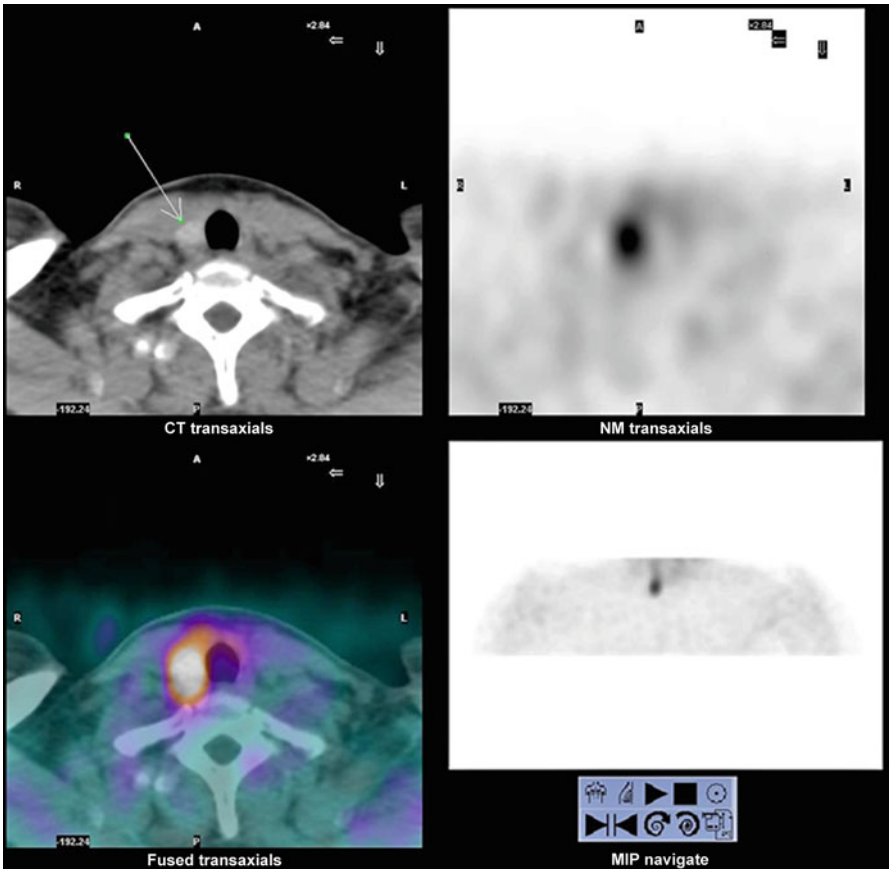


Fig. 8.7 Parathyroid scan. Two-hour image set from a dual time point MIBI SPECT/CT study. Intense retained uptake is shown in a right superior parathyroid adenoma; CT and fusion images show the adenoma (*arrow*) as a hyperdense nodule posteromedial to the thyroid. (*Left upper*) SPECT image; (*left lower*) MIP image; (*right upper*) CT image; (*right lower*) fusion SPECT/CT

concentrated on three tracers: conventional bone scanning – with or without SPECT – which is dealt with elsewhere; radiolabeled white cell scanning; and ^{18}F-FDG PET, which again lies outside the scope of this chapter.

White cell scanning requires the separation of leukocytes from a sample of fresh blood, in vitro labeling, and reinjection and therefore requires the facilities for handling of blood products in appropriate sterile environments. The cells can be labeled with ^{111}indium but more commonly are labeled with ^{99m}Tc HMPAO.

In clinical practice, HMPAO white cell scanning is most commonly employed to detect orthopedic infections, either as primary osteomyelitis or increasingly to assess suspected infection in large joint prostheses. For such indications, planar imaging of limb bones is expected to yield sufficiently accurate results, with SPECT rarely adding significant value. Equally, the deleterious effect of attenuation artifact

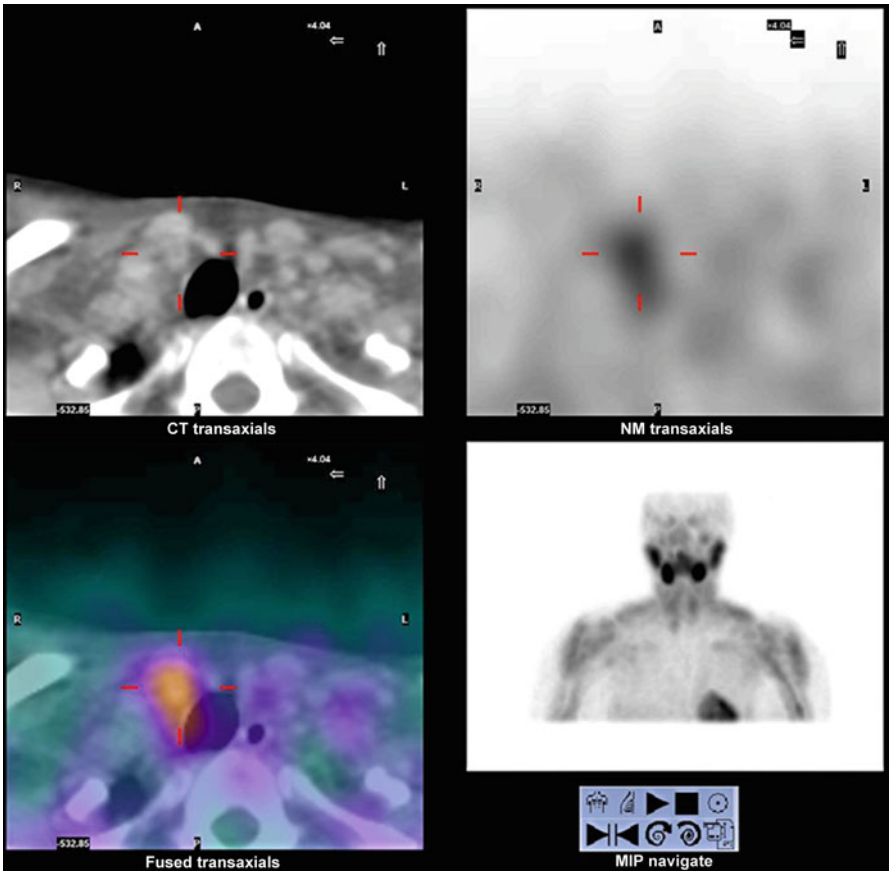


Fig. 8.8 Parathyroid scan. Two-hour image set from a dual time point MIBI SPECT/CT study. Retained uptake is shown in a small right inferior parathyroid adenoma (crosshairs); CT images show the adenoma as a hyperdense nodule inferior to the thyroid. (*Left upper*) SPECT image; (*left lower*) MIP image; (*right upper*) CT image; (*right lower*) fusion SPECT/CT

on CT images from metal-bone interfaces would be expected to significantly limit the clinical utility of CT. However, in a recent paper [33] looking at the assessment of suspected infection in hip and knee prostheses, SPECT was shown to give a significant incremental value over planar imaging, with sensitivity increased from 66 to 89 % and negative predictive value from 81 to 91 %. The addition of SPECT/CT showed a further small increase in negative predictive value to 94 %, with increased diagnostic confidence in the localization of low-grade periprosthetic infections.

White cell scanning is also advocated for the assessment of active inflammatory bowel disease, although this indication has largely been supplanted by increasing use of small bowel MRI scanning. However, there remains a niche role for scintigraphy in assessing disease activity in difficult patients with known complex disease

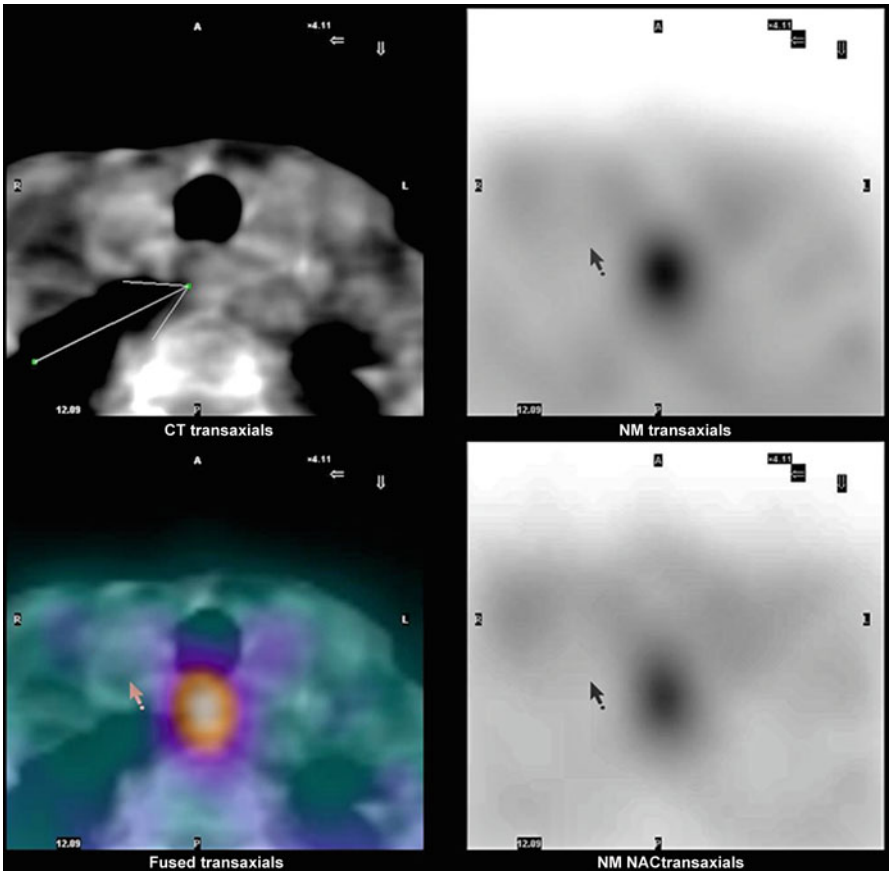


Fig. 8.9 Parathyroid scan. Two-hour image set from a dual time point MIBI SPECT/CT study. Retained uptake is shown in an ectopic parathyroid adenoma; CT images show the adenoma immediately posterior to the upper trachea and adjacent to the esophagus (*arrow*), allowing preoperative anatomical localization not possible on ultrasound. (*Left upper*) SPECT image; (*left lower*) MIP image; (*right upper*) CT image; (*right lower*) fusion SPECT/CT

due to long-standing Crohn's disease or extensive previous surgery [34]; in these patients, accurate localization of activity may be difficult, and the additional value of SPECT/CT will mirror the impact seen in tumor imaging with octreotide. Similar results may also be achieved with FDG PET/CT.

In vascular surgery as in orthopedic surgery, infection may complicate graft insertion and lead to significant complications and morbidity. Although rare, occurring in between 0.5 and 3 % of cases, the risks of infection include limb loss and death in up to 40 % of cases [35]. White cell scanning has been shown to detect periprosthetic infection in patients with negative or equivocal CT results but with variable reported sensitivity. This in part may be due to the difficulty in separating abnormal WBC accumulation from normal vascular blood pool. A recent paper

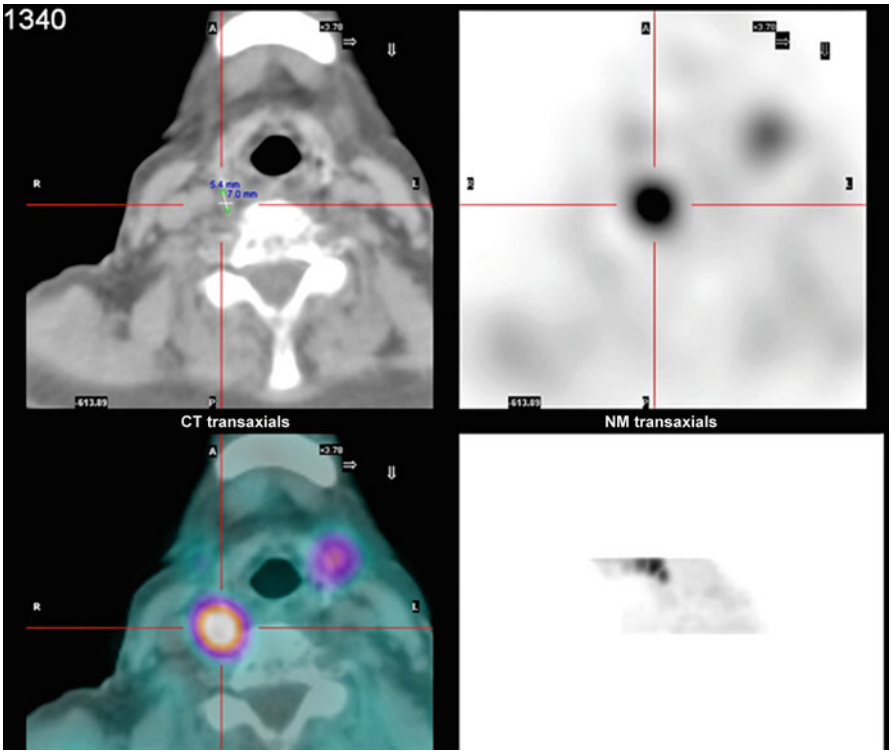


Fig. 8.10 Parathyroid scan. Two-hour image set from a dual time point MIBI SPECT/CT study. Retained uptake is shown in a small right ectopic parathyroid adenoma (crosshairs); SPECT/CT images show the adenoma lying superior to the thyroid and at the level of the hyoid bone. (*Left upper*) SPECT image; (*left lower*) MIP image; (*right upper*) CT image; (*right lower*) fusion SPECT/CT

analyzed the results of SPECT/CT imaging [36] and, although a relatively small series, showed a much higher accuracy for this technique in localizing suspected periprosthetic infection in arterial grafts.

Finally, white cell scanning has been used to evaluate patients with pyrexia of unknown origin (PUO). However, the incidence of bacterial infections as a cause for PUO is relatively rare, and for this reason, reported results for WBC scanning are disappointing [37]; while SPECT/CT may improve localization, this is unlikely to alter this limitation and thus will not supplant FDG PET/CT in this role.

In summary, for SPECT/CT in oncology, endocrine imaging and infection, because of its ability to harness the anatomical localization of CT with the functional information of nuclear medicine, is likely to improve diagnostic utility in a number of indications. In oncology, this will include tumor-specific imaging agents, and in clinical practice in the UK, this is likely to be in octreotide imaging of neuroendocrine tumors, iodine imaging of suspected thyroid carcinoma

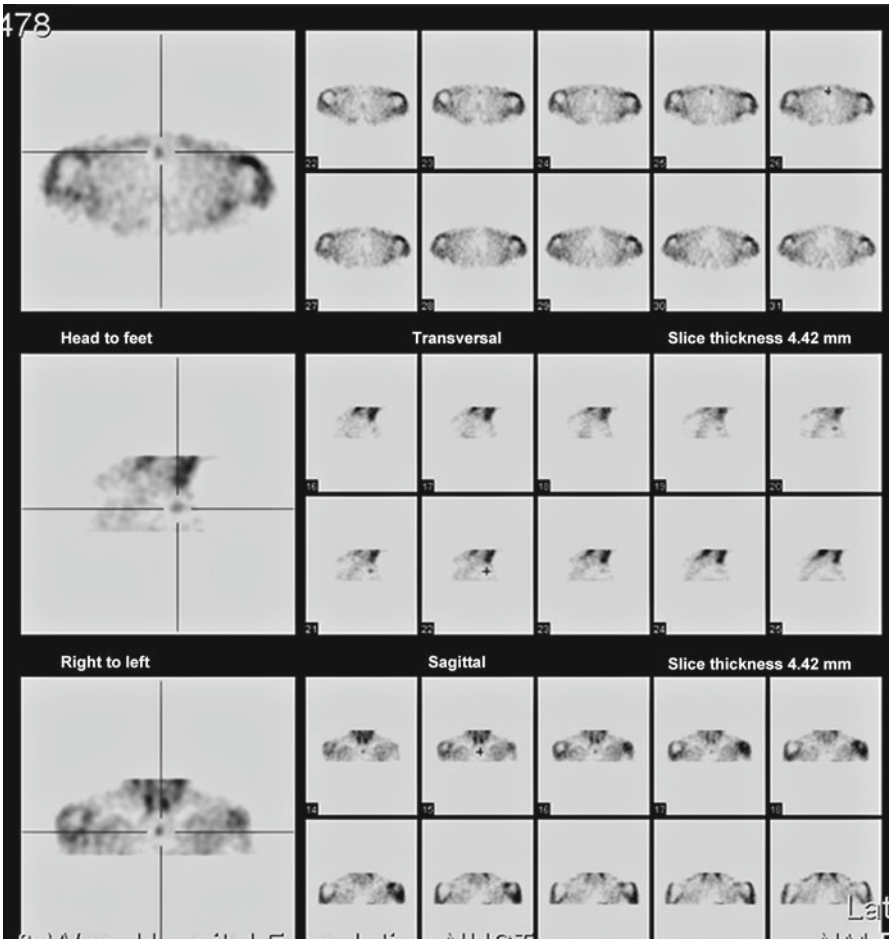
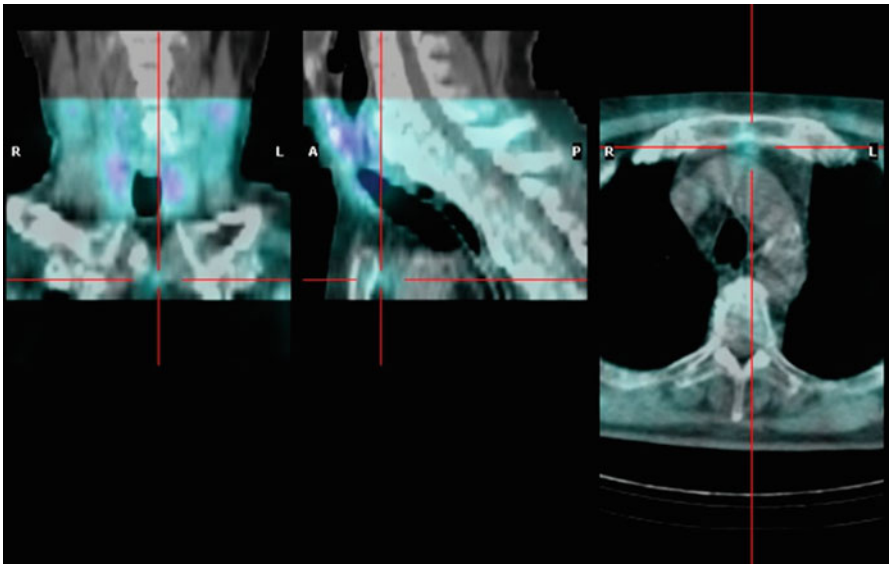


Fig. 8.11 Parathyroid scan. *Image 1*: Triangulated 3 plane image set from SPECT study at 2-h post-MIBI injection showing small left parathyroid adenoma lying below root of neck

recurrence, and possibly MIBG imaging of pheochromocytoma. SPECT/CT will also be indicated for sentinel node localization in melanomas, in pelvic cancers, and potentially in difficult case of breast cancer. In endocrine imaging, in addition to MIBG, SPECT/CT is indicated in the localization of ectopic parathyroid adenoma using MIBI and may have a role in localizing normally located parathyroid lesions as a single test replacing ultrasound prior to minimally invasive surgery. Finally, SPECT/CT will in selected cases give added value in white cell scanning, particularly within the chest or abdomen or in the management of infected vascular grafts.



- Image 2
- SPECT CT image set showing uptake in a 6mm nodule in the anterior mediastinal fat, anterior to the aortic arch

Fig. 8.12 Parathyroid scan. *Image 2*: SPECT/CT image set showing uptake in a 6 mm nodule in the anterior mediastinal fat, anterior to the aortic arch

8.9 Imaging the Heart Using SPECT/CT

8.9.1 Introduction

Myocardial perfusion imaging (MPI) is now well established for assessing coronary artery disease. MPI is one of the most widely used methods for functional assessment of coronary artery disease with more than eight million patients undergoing this investigation in the USA alone [38]. It is also widely used in the UK and Europe. Inclusion of MPI in coronary artery disease investigation is shown to be cost-effective and has been recommended in the National Institute for Clinical Excellence (NICE) guidelines in the UK for these patients [39, 40]. There has been continuous development in the imaging hardware, software, and interpretation of the images. One of the major problems has been the tissue attenuation, resulting in artifactual areas of apparent abnormal perfusion which are due to tissue attenuation from breast and other structures in the chest.

An immense asset in increasing the accuracy in reporting and minimizing misinterpretation of these artifacts is the use of normal libraries specific to each gender. This

has greatly increased the confidence in reporting. The limiting factor in using normal libraries is the variations in the attenuation in individuals as the breast size and shape and heart position in the mediastinum is different between individuals and depends on body stature and habitués. Average normal libraries do not always tally with the individual patient variations. This may lead to erroneous interpretation of the artifacts with possible over or under diagnosis of significant coronary artery disease.

The introduction of hybrid cameras combining the single photon emission computed tomography with X-ray computerized tomography (SPECT/CT) has enabled more accurate measurement of relative blood flow in the myocardium, using a low-dose CT for accurate attenuation correction of the images for each patient. SPECT/CT attenuation correction has enabled a more confident and accurate interpretation of myocardial perfusion images [41–45]. Many departments now routinely use attenuation-corrected images for more confident reporting. Attenuation-corrected normal libraries available for various commercial cardiac reporting packages have further increased confidence in accurate diagnosis and quantification of ischemia. Initially, SPECT/CT hybrid cameras had a low-dose nondiagnostic CT for attenuation correction only. Now most of the modern hybrid cameras incorporate high-quality diagnostic CT scanners. With these cameras calcium scoring, MPI and even coronary CT angiography can be performed on the same imaging system. This has enabled some departments to entertain one-stop comprehensive diagnostic investigation of patients with chest pain, by combining calcium scoring and functional and structural imaging in one session. Hybrid imaging has also enabled a better localization and assessment of the significant structural lesions. With PET/CT cameras, absolute myocardial flow quantification is possible; however, they are not at present as widely available.

8.9.2 A Clinical Perspective on Myocardial Image Artifacts

Gamma rays are absorbed and scattered while passing through various tissues in the body, before they reach the camera detectors. The extent of this loss and scatter depends on the energy of the gamma rays, the tissue density, and the amount of tissue they have to traverse before reaching the detectors. The tissue density (air/water/bone) varies in different areas and directions. Nonuniform degradation of the gamma rays results in fewer gamma rays reaching the camera detectors depending on the angle of acquisition, degrading the quality of images.

The extent of attenuation is different between individuals due to differences in body shape, weight, breast shape and size, position and thickness of diaphragm, and position of heart in the mediastinum. The most common attenuation artifacts are due to attenuation from breast soft tissue (Fig. 8.13) being most pronounced in women and diaphragmatic attenuation artifact (Fig. 8.14) in men due to position of the heart behind the dome of diaphragm and respiratory motion of the diaphragm. Interpretation of images can be even more complicated as diaphragmatic attenuation can be seen in women and breast or chest wall attenuation in larger men. Breast attenuation is seen as relative reduction in the anterior and/or anteroseptal wall counts (Fig. 8.15), and if

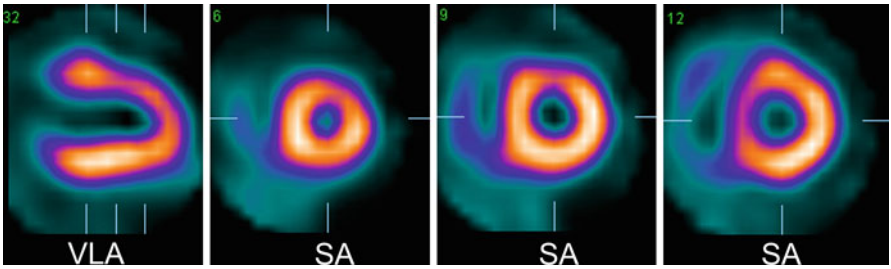


Fig. 8.13 Typical anterior attenuation artifact produced by breast attenuation. The artifact is clearly seen in the vertical long-axis (*VLA*) and in the apical and mid short-axis (*SA*) cuts in the non-attenuation-corrected perfusion images

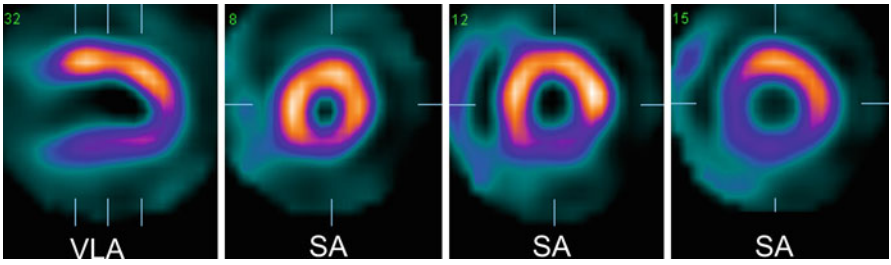


Fig. 8.14 Inferior wall artifact typically seen in men due to diaphragmatic attenuation. This is clearly seen in the inferior wall of vertical long-axis (*VAL*) and the mid and basal short-axis (*SA*) cuts in non-attenuation-corrected images

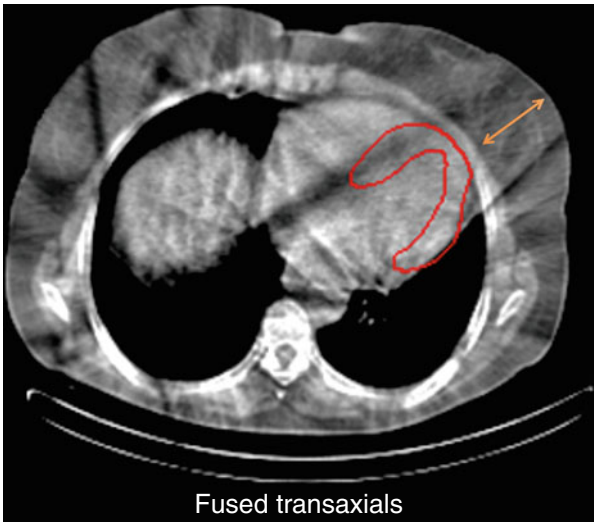


Fig. 8.15 Fused transaxial perfusion and CT images. The perfusion emission has to transverse through varying thickness in chest wall including the breast tissue (*arrow*) introducing artifacts into emission images

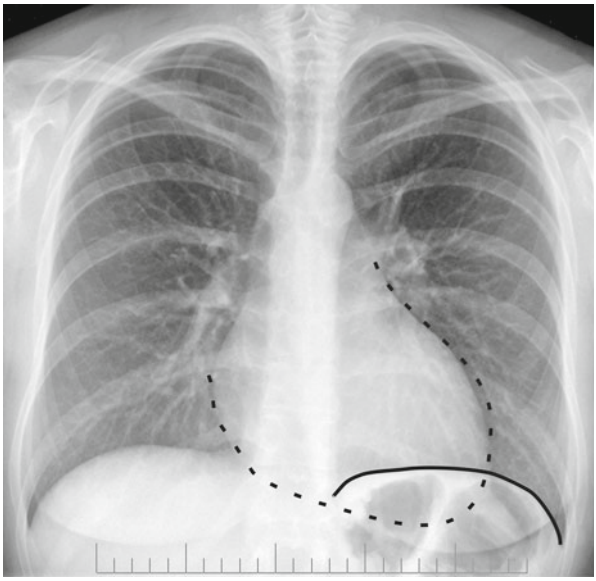


Fig. 8.16 Heart (*dotted line*) is positioned on the diaphragm, but part of it is hidden behind the dome of diaphragm (*solid line*). The position of the heart and its relation to diaphragm vary from person to person and can attenuate the emission data to varying degrees resulting in attenuation artifacts in reconstructed images

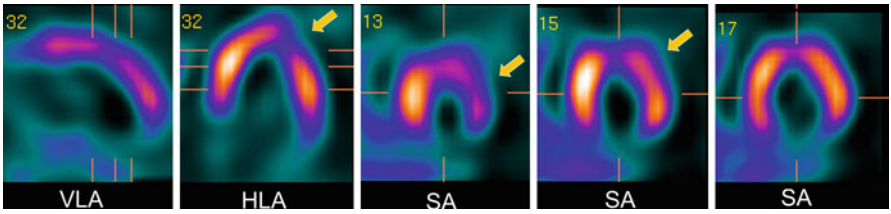


Fig. 8.17 This patient was unable to raise arm above head for perfusion imaging. The emission gamma rays have to traverse the arm to varying degrees in the lateral projection images. This has resulted in an area of apparent reduced perfusion in the anterolateral wall near the apex best seen in horizontal long-axis (*HLA*) and apical short-axis (*SA*) cuts. There is also a true perfusion defect in the inferior wall in this patient seen in vertical long-axis (*VLA*) and short-axis cuts

the breast is positioned differently during rest and stress acquisitions, this may lead to apparent reversible ischemia which is not an ischemic lesion. Diaphragmatic attenuation artifact typically affects the inferior wall (Fig. 8.16). The extent and severity of the defect can vary considerably. In areas with known attenuation artifacts, assessing for ischemia can be difficult. Other sources of attenuation include prosthetic breast and arm attenuation. Patients are ideally imaged supine with the arms positioned over the head out of the cardiac field of view. If the patient is unable to raise the arms (usually due to arthritis), then the arm which lies in the field of view would introduce further attenuation artifacts which are usually seen in the anterolateral wall (Fig. 8.17). For consistency, the patient should be imaged with the arm in the same position for rest and stress, to enable easier comparison of the two scans.

There are several suggested techniques to assess for attenuation artifacts in the absence of attenuation maps. The most common method is to compare the images to a normal library image which is gender specific and takes into consideration the aver-

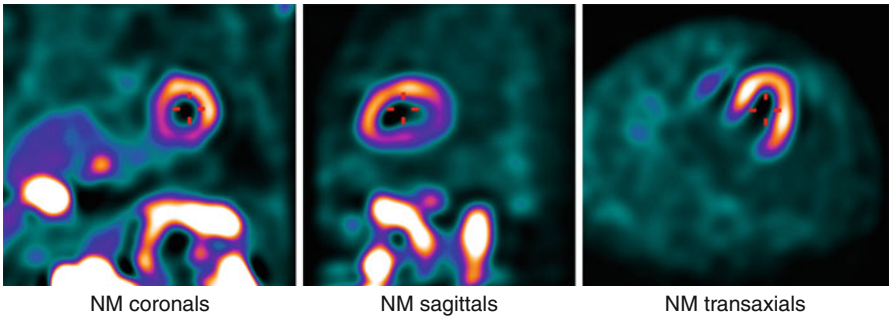


Fig. 8.18 Coronal, sagittal, and transaxial emission reconstructed tomography images showing the distribution of the radiopharmaceutical

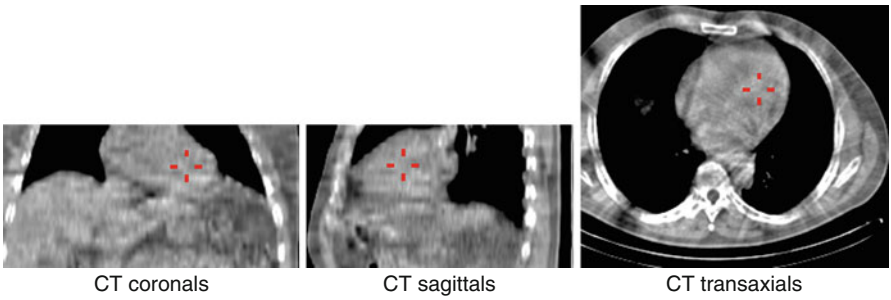


Fig. 8.19 Transmission CT images delineate different density tissues used for attenuation correction of the emission data

age reduction in counts. As mentioned earlier, unfortunately not all patients are average, and at best, this is an approximation. Another suggested method is imaging the patient in a prone position, which may reduce the artifacts. However, at least in theory, the best method to reduce attenuation artifacts is using a SPECT/CT system which can correct for attenuation using a transmission image for each individual patient.

8.10 Correcting for Attenuation Artifacts Using CT

Myocardial perfusion images are produced by reconstruction of acquired emission data, emitted from radiopharmaceutical trapped in the myocardium proportional to myocardial blood flow (Fig. 8.18). For attenuation correction, an external radiation source is used to produce transmission maps which show the body contour and the tissue composition (air/water/bone). This image is used for correcting the amount attenuation of the emitted gamma rays as they pass through these tissues with different densities (Fig. 8.19). The early cameras used a gadolinium-153 external source attached to the head of a camera. Gadolinium has a relatively short half-life and needs to be replaced several times during the life of a gamma camera and can be

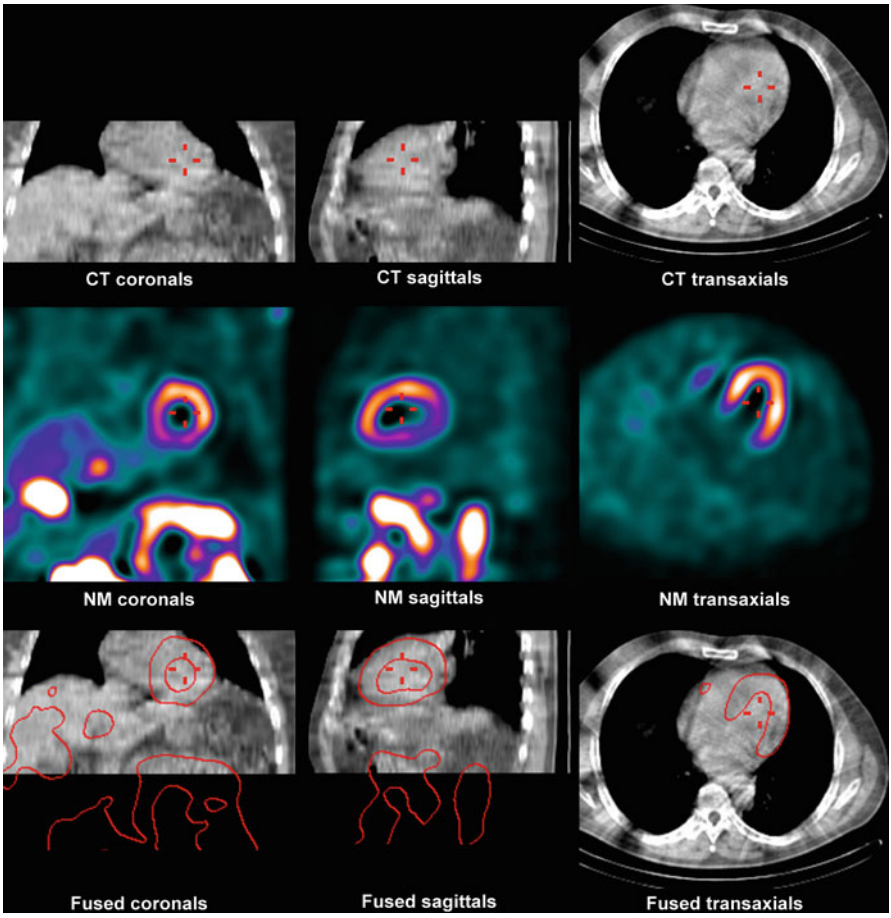


Fig. 8.20 There is good co-registration of CT (*top row*) and emission (*middle row*) data. Co-registration is best assessed by superimposing the CT and emission data (*bottom row*)

expensive. Other longer life sources have also been used such as barium-133. CT has gradually replaced these systems, as it is more convenient, readily available, well tested, and can produce additional information for accurate localization and diagnosis. SPECT/CT is now the most commonly used method for attenuation correction. A standard CT scanner is attached to the back of the gamma camera. The SPECT images are acquired first. Then the camera bed is automatically moved to the correct position in the CT scanner to acquire corresponding attenuation or diagnostic CT images. Precise positioning of the imaging table enables easier and more accurate co-registration of images. The CT scanners vary in quality and speed. It can be a slow 5-min acquisition or very fast multislice CT acquisition. For attenuation correction per se, there is no need for high-quality CT images as the nuclear medicine images themselves are of relatively low resolution. Whichever system is used, attenuation

correction protocols deliver lower radiation dose to the patient when compared with diagnostic CT imaging. With the introduction of fast CT scanners, imaging may be performed at a specific respiratory phase, while the emission images are produced over an averaged respiratory phase. This may result in misregistration. Various techniques such as shallow-breathing CT or respiratory-averaged CT are employed to overcome this problem. Different manufacturers employ different algorithms and methods to co-register emission and transmission images. Automated techniques include mutual information, edge detection, and radon consistency [46–48]. Accurate co-registration of the images is essential (Fig. 8.20). Once attenuation correction maps are produced and co-registered with the transmission images, then the emission data are corrected for the tissue and the distance they travel from the emission source to the camera head. This can produce images which are independent of attenuation and thus avoiding the artifacts associated with attenuation.

8.11 Quality Checks and Avoiding Pitfalls

Single photon emission tomographic images for myocardial perfusion images are normally acquired in a 180° acquisition from left posterior oblique to right anterior oblique. The length of the acquisition depends on the pharmaceutical, the amount of injected activity and camera/collimator sensitivity, and the processing parameters used (e.g., resolution recovery). A typical study takes 10–20 min. During acquisition, the patient may move, or the position of the heart may change as patient relaxes (known as cardiac creep). The study should be checked for movement at the end of the acquisition by playing a cine display of the raw emission images. Various automated and manual software packages are available for movement correction. It is important to check for movement immediately after the acquisition before the patient is sent home. If the movement is excessive, the acquisition may have to be repeated. It should be emphasized that only the position of heart should be considered for movement and not other areas in the image and the activity in the gut moves during the acquisition time. Movement correction is achieved by moving the entire images to align the heart in successive frames. Movement along the y-axis can be corrected. Motion along the x-axis is more difficult to correct, and correcting for x-axis motion may itself introduce artifacts, and it is best to avoid by default. Motion may also occur during the acquisition of a single view. This will result in blurred images and of course cannot be corrected. If there is excessive motion, a repeat study may be required. It is also best to check for motion, and the need for repeat study immediately after emission acquisition before CT is performed. If the study has to be repeated, the CT acquisition can then be aborted and performed only at the end of the repeat study to avoid unnecessary radiation to the patient. As the CT transmission images are acquired after the completion of the SPECT images, the heart on CT images will correspond to the position of heart in the final projection of the MPI study. This may be a different position from the early frame of the SPECT

acquisition and may be a problem especially if movement correction has been applied. The image may have moved to a different position by motion correction from that of the final position of heart when the CT images were acquired. Cardiac creep is slow movement of the heart during acquisition probably due to patient relaxation. The typical artifact is a jump in the images, as the last image from head one is stitched to the first image of the head two. As the patient relaxes, the heart usually moves inferiorly in the chest. At the end of the study, the heart is at lowest position when the CT is acquired. Motion correction for creep may push all the SPECT image data to the original (beginning) frame when the heart position was higher in the chest, and the SPECT and CT data may be misregistered. SPECT/CT misregistration may also occur in cameras systems with very fast CTs. The CT images may be acquired in a certain respiratory position of the heart, while the SPECT images are averaged respiratory position over the period of acquisition and are independent of respiratory movement. Co-registration of fast CT images and SPECT images may be more difficult. Several methods such as shallow-breathing or respiratory-averaged CT have been proposed to reduce this kind of misregistration. In an ideal patient, no movement or cardiac creep will have occurred; here there is good co-registration of SPECT and CT data resulting in accurate attenuation correction. However, because of the problems mentioned, it is imperative that the registration of SPECT and CT data is assessed before attenuation correction is performed. Software tools should be available in SPECT/CT systems for this purpose. It should be possible to see the transmission (attenuation) map and SPECT images separately and then be able to check registration by superimposing the two sets of images. CT data is displayed in gray scale with superimposed SPECT images in color scale. The registration should be checked in all three transaxial, sagittal, and coronal cuts. Figure 8.20 shows good registration between SPECT and CT resulting in accurate attenuation correction. Figure 8.21 shows reconstructed perfusion images (SPECT) in a male patient. Note the normalization of the inferior wall diaphragm attenuation artifact with attenuation correction. Figure 8.22 shows misregistration between CT and MPI images, where MPI images are positioned inferiorly in the chest in comparison to CT images. This may result in introduction of artifacts which may lead to misinterpretation of the MPI study. In this study, misregistration resulted in an apparent anterior defect on attenuation-corrected images only. Many software systems nowadays allow realignment of the two set of data. Figure 8.23 shows artifact caused by misregistration which is corrected following correct realignment of the SPECT/CT registration. The correction may not be possible if there is excessive patient motion during the study. The reporter should be aware of any misregistration and/or attempted corrections. The attenuation-corrected images should then be reported cautiously.

A possible artifact following attenuation correction in some patients is relative hypoperfusion of the anterior wall. This in particular can be a problem in patients with high gut and liver activity. The gut activity lies more or less central in the body and increases most with attenuation correction relative to the rest of the image. In addition, gut activity moves during the study and can have different position in different raw image frames and thus leading to reconstruction artifacts. High gut activity in the close vicinity of the heart may cause spurious apparent high counts in the

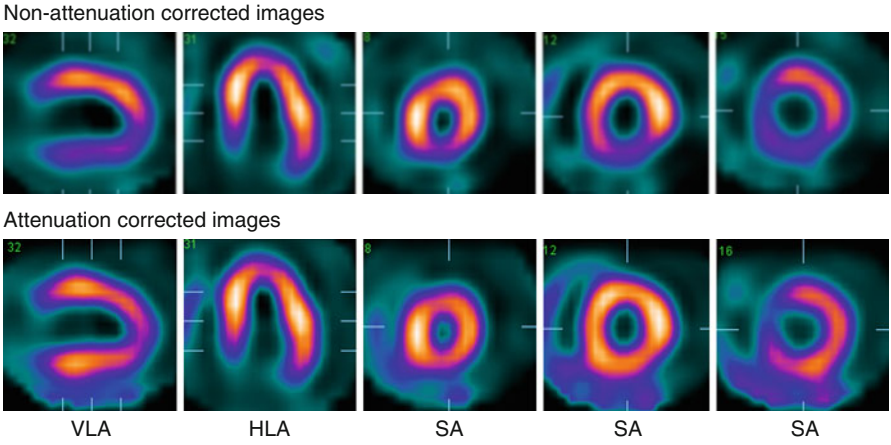


Fig. 8.21 Reduced counts in the inferior wall are due to attenuation artifact. With attenuation correction, the artifact is removed and there is more uniform distribution of counts in the myocardium

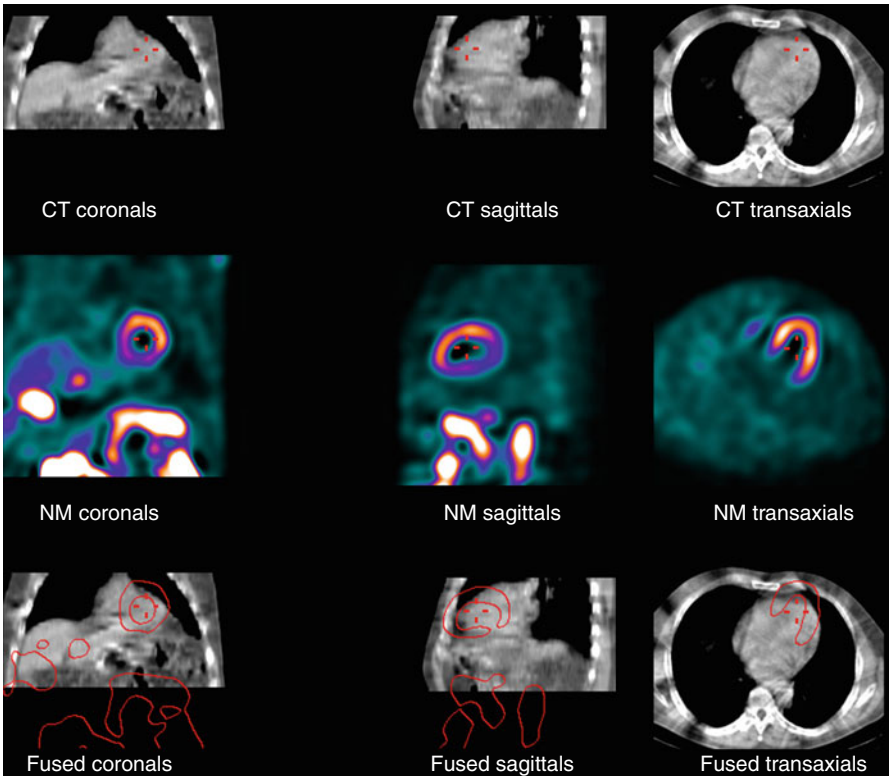


Fig. 8.22 Images should be checked for good registration before applying attenuation correction. This is an example of misregistration where the emission data are positioned anterior to the CT images. This should be corrected before attenuation correction is applied to the emission data, to avoid introducing artifacts to the perfusion images

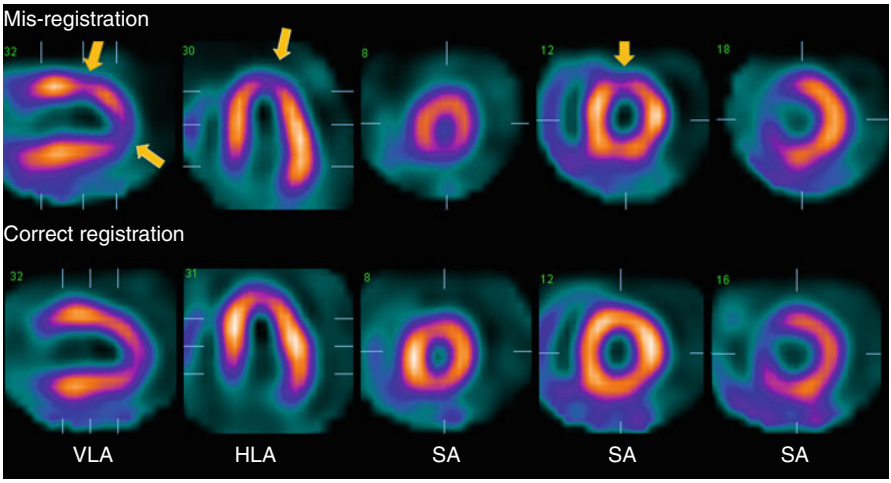


Fig. 8.23 Anterior and apical artifacts produced by misregistration of emission and transmission images (*top row*). When registration is corrected, the artifacts are no longer seen (*bottom row*). *VLA* vertical long axis, *HLA* horizontal long axis, *SA* short axis

inferior (posterior) wall of the myocardium. With image normalization between stress and rest study, the high activity in the inferior wall may result in apparent reduction in the anterior wall activity and, if this happens at stress only, then may lead to an apparent anterior wall reversible defect. Anterior wall in attenuation-corrected images with high gut activity should be viewed cautiously.

As mentioned above, different manufacturers use different transmission data and different algorithms for attenuation correction. One should be aware of differences in the resulting attenuation-corrected images [45], and in particular if a department is using several different systems, the reporter should be aware of these and take the differences into consideration during reporting.

8.12 Using Attenuation-Corrected Images for Reporting

Use of attenuation-corrected images has been controversial during the past decade; however, there is increasing evidence that in conjunction with standard non-attenuation-corrected images, it can improve the confidence and accuracy of reporting [46]. It is now generally recommended to be used for more accurate reporting of MPI [42–44]. When reporting, it is best to start with the normalized standard (non-attenuation-corrected) stress and rest images first, as most people are most familiar with these images. Once potential ischemic areas are noted, one should look at the attenuation-corrected images. It will increase the confidence in detecting an ischemic lesion if the lesions tally with the standard images. If the lesions seen in standard images are no longer visible, one should reconsider the possibility of artifactual lesions and review the standard images as the attenuation-corrected images are

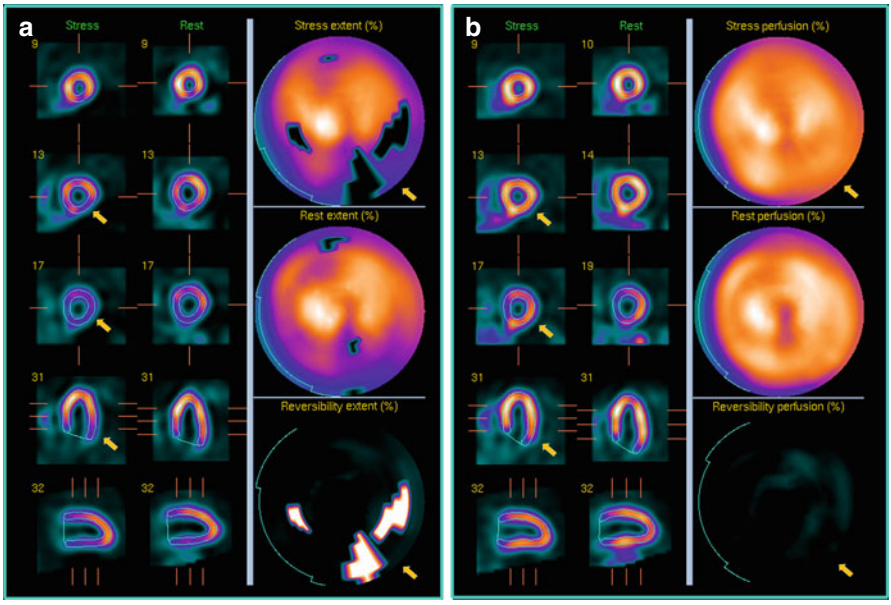


Fig. 8.24 Perfusion images without attenuation correction (**a**) show an area of reversible ischemia in the inferolateral wall. Attenuation-corrected images correct for this lesion and indicate no ischemic areas (**b**). Apparent areas of reversible ischemia in non-attenuation-corrected images may be produced by different positioning of the attenuating soft tissue at rest and stress

increasingly recognized to be more accurate provided the pitfalls are avoided. Myocardial display packages should be used for normalization of rest against stress images. They also supply additional information by providing various polar map displays and semiquantification (extent and severity of defect, scoring system) and by comparing to a set of normal libraries. Often the attenuation-corrected images tally with the standard images corrected for normal libraries. However, normal libraries correct for average artifacts. The bull's-eye images may overestimate or underestimate the extent of a lesion. It is important to consider standard attenuation correction and bull's-eye display (with normal library comparison) for more accurate and confident reporting. If standard and attenuation-corrected images both show a defect but are not picked up in bull's-eye display, it is likely that the bull's-eye images are missing a true lesion. As the methodology has improved and confidence in attenuation correction has grown, it is realized that often in patients with, for example, larger than average breast, the attenuation-corrected images accurately show a normal distribution, while bull's-eye and standard images may display a defect which does not exist. Figure 8.24 is an example of such patient where the inferior wall defect seen both in standard images and bull's-eye display of standard images strongly indicated true ischemia, but the attenuation-corrected images were normal. This patient underwent angiography – which showed normal coronary arteries (Figure 8.25). Occurrence of a defect in attenuation-corrected images which was not present in standard images is more difficult to assess, and reviewing co-registration or looking for other causes

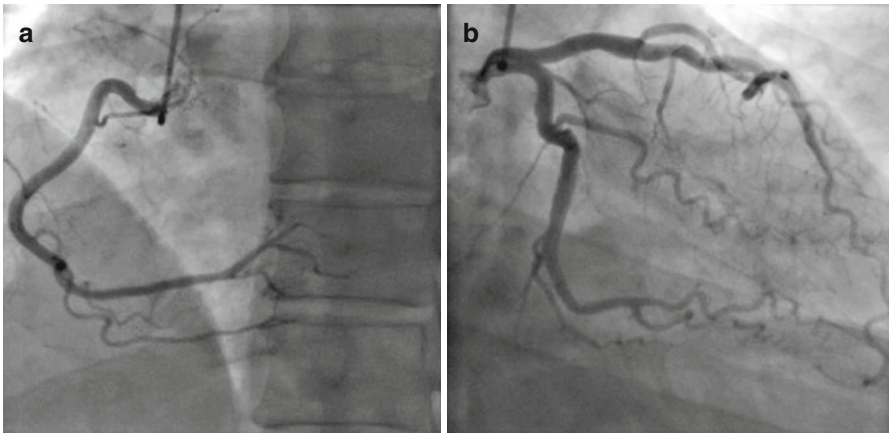


Fig. 8.25 Coronary artery shows no significant lesion in the right (**a**) or the left (**b**) coronary arteries

such as gut activity is essential. Unfortunately due to variation in the attenuation correction techniques and algorithm, it is not as yet recommended to only rely on attenuation-corrected images; review and reliance on standard images is still considered essential for accurate reporting.

There are now a growing number of software packages which also include normal libraries for attenuation-corrected images. The use of normal libraries although very helpful for standard images, where there is significant attenuation artifact, is less useful in attenuation-corrected images as one would expect less artifacts. However, they can be useful for more accurate assessment of the extent and severity of the perfusion defects.

8.12.1 Hybrid Imaging

SPECT/CT hybrid cameras have developed considerably from the early days when a low-specification CT was simply bolted on the back of a standard SPECT camera. Most of the modern SPECT/CT cameras now have diagnostic CT capability. These CT systems are able to deliver high-quality CT images. High-quality CT imaging has also broadened the scope of utilization of hybrid cameras for cardiac use. In the NICE guidelines for management of acute chest pain [40], calcium scoring is now recommended as the initial diagnostic investigation in patients with low probability of ischemic heart disease presenting for the first time. Modern SPECT/CT cameras are increasingly used for effective and streamlined management of such patients. Calcium scoring can now be performed in patients referred from acute chest pain clinics before performing a perfusion scan, and combined with clinical probability, a decision can be made to continue and go ahead to myocardial perfusion scanning on the same imaging session. This is already implemented in some departments. There are now also several

64 slice SPECT/CT scanners available. With these hybrid scanners, it is possible to perform coronary artery CT angiography on the same session for these patients. An example of a department with a one-stop diagnostic imaging protocol for acute chest pain is where the patient is referred from chest pain clinic, depending on the clinical probability; calcium scoring can be performed first; some patients may need no further investigation, while others can proceed directly to the functional imaging (myocardial perfusion scanning), if normal may not require further investigation; and others may need to proceed further to coronary CT angiography. This equipment is fairly new, and the best protocols for these cameras are still being considered. Some of these hybrid cameras incorporating new hardware (solid-state cameras, multiple pinhole, or cardiofocal collimators) and new software (resolution recovery) have decreased the perfusion imaging time (e.g., 5-min imaging) to match those of other modalities. Fast and effective diagnostic clinical pathways can be devised using these cameras to optimize management of patients with ischemic heart disease.

Hybrid cameras have also enabled routine use of myocardial perfusion images and coronary CT angiography images for better delineation of exact coronary artery disease causing significant ischemia. With hybrid coronary angiography and SPECT images, it is possible to accurately determine the significance of arterial narrowing for better targeting for treatment. There is growing evidence that hybrid imaging is superior to either myocardial perfusion imaging or coronary CT angiography alone [45, 49]. The joint positional statement of European Association of Nuclear Medicine and European Council of Nuclear Cardiology states the available evidence to support the advantage of hybrid imaging [50].

8.13 An Overview of the Use of SPECT/CT in Skeletal Disease

Skeletal scintigraphy (bone scan) has a long history in the nuclear imaging arsenal, first being used some 35 years ago. Skeletal scintigraphy offers advantages over conventional imaging techniques, giving a whole body overview with an acceptable radiation dose of around 5 mSv, defining the extent of bony involvement through the demonstration of altered bone turnover [51, 52]. This gives the study great sensitivity, but the images lack specificity.

SPECT techniques increase lesion visualization and increase lesion localization but remain nonspecific. Hybrid scanners allow the co-acquisition of the radionuclide dataset with that of a CT dataset, allowing for more accurate lesion localization and characterization through the CT component. This is particularly true with HP-CT systems that combine nuclear imaging with a fully diagnostic CT scanning ability. The ability of fused SPECT/CT to reach a definitive diagnosis combines the sensitivity of the bone scan with the diagnostic certainty added by contemporaneous CT.

The use of diagnostic CT does inevitably have a radiation dose penalty to the patient, and its use needs to be considered and to be of benefit to the patient. Consideration needs to be given to the additional diagnostic information that can be gained from the study. A trade-off can be made, accepting less diagnostic images with a lower dose when localization is of greater importance than characterization.

HP-CT is of less benefit in those with a normal planar or SPECT study or those with a pattern obvious for disseminated bony metastases or other pathology. Previous imaging of the patient may obviate its need, that is, a recent diagnostic CT or MRI that covers the area of scintigraphic abnormality. Conventional radiographs should suffice for indeterminate peripheral/extremity lesions. HP-CT is of benefit in assessing indeterminate lesions of the axial skeleton and in particular the spine. It is also useful in areas that conventional radiography interpretation can be difficult due to overlap/overlay of small structures such as the carpus and tarsus and also in characterization of abnormalities in and around the base of the skull.

A number of papers attest as to the improved sensitivity and specificity by adding SPECT and/or SPECT/CT to the standard bone scan. Romer et al. [53] demonstrated the usefulness of SPECT/CT in evaluating foci of increased bone metabolism that were classified as indeterminate on SPECT in cancer patients. In this study, they showed increased performance in lesion characterization usefully differentiating physiologically active benign lesions from active metastatic deposits. This largely revolved around spondyloarthropathic changes in the vertebrae and facet joints. Using SPECT-guided CT, the authors were able to confidently classify 92 % of indeterminate lesions, thus improving the value of the examination itself and avoiding the need for additional, often expensive correlating studies such as MRI. Of note is the distribution of the indeterminate findings. The majority were within the spine itself with the minority in the pelvis, scapula, sternum, sternoclavicular joint, and ribs. Only the rib lesions remained indeterminate following SPECT/CT.

Subsequently, a paper by Strobel et al. [54] evaluated the performance of planar imaging against SPECT imaging and against SPECT fused with CT for lesions within the axial skeleton. Lesion visibility, diagnostic performance, and confidence levels in the diagnosis were assessed. The reported sensitivities for the three methods were 82, 91, and 100 % with specificities of 94, 94, and 100 %, respectively. The study had relatively low numbers, and these figures were not demonstrable as a statistically significant difference between the three methods, but reporter confidence and reporting certainty were higher for the fused datasets over the more conventional planar or SPECT images. This increased confidence in interpretation is an important factor and has been demonstrated in other studies [55].

Bone scintigraphy is well established and has long been used in assessing skeletal disease dissemination in known malignancies due to having superior sensitivity over radiographs. Skeletal lesions are visualized on plain radiography when there has been loss of approximately 50 % of the cortex. Conversely, the skeletal scintigram is positive when only 5 % of the cortex is involved [51, 52]. These figures are tumor dependant. The bone scan is of proven benefit in prostate cancer [56] and is often useful in breast and lung cancer although less routinely. It is less useful in those cancers that have a more infiltrative pattern of dissemination [57] such as lymphoma and has been superseded by PET scanning in those tumors. The bone scan is not routinely indicated in myeloma; often significantly underestimating disease and skeletal surveys is more routinely utilized for this condition. Purely lytic lesions such as can be seen in renal tumors, for example, are more difficult to iden-

tify with bone scan as they are often photopenic but increasing use of SPECT and SPECT/CT improves small lesion detection.

The purpose of imaging in cancer is to give accurate staging information. This in turn informs diagnosis and treatment options and allows for evaluation of treatment response and prognosis. Accurate lesion detection and characterization is therefore of fundamental importance. Metastases in the spine are common [58] as are degenerative conditions, and SPECT/CT offers improved certainty in diagnosis over planar or SPECT images alone. Strobel et al. [54] and Utsunomiya et al. [55] demonstrated increased accuracy and confidence in diagnosis when utilizing SPECT/CT to characterize indeterminate lesions of the spine in cancer patients. A further paper on the imaging of malignant bone involvement by Even-Sapir [59] confirms the usefulness of hybrid techniques. This is achieved in part by overcoming superimposition of structures on planar imaging with the superior localization of lesions afforded by SPECT and SPECT/CT techniques. Accurate localization of lesions improves specificity as metastases tend to affect characteristic sites such as the vertebral pedicles; see Fig. 8.26.

The ability to proceed to perform a staging CT at the same visit if an incidental finding is made of a metastatic bone lesion is useful in the diagnostic end CT systems and can save the patient a further appointment, potentially shortening their pathway. The hybrid imaging capabilities allow for the development of one-stop imaging pathways which is something we routinely utilize in staging breast cancer [60].

Conventional radiographs can be used for the assessment of indeterminate lesions to correlate peripheral or extremity findings on the planar or SPECT study and utilize SPECT/CT in axial and pelvic findings particularly when these are monostotic. Polyostotic disease is usually due to disseminated disease, but CT can be useful if there is any doubt, that is, second lesion outside of radiotherapy field which would preclude a radical approach with radiotherapy.

Primary bone malignancies can be staged with MRI due to the superior soft tissue discrimination, but the bone scan is of use in assessing polyostotic or metastatic disease. SPECT/CT can also be useful in guiding bone biopsies, the radionuclide element showing sites of active disease and the CT component allowing for planning of surgical or percutaneous approach.

The CT scanning component of the hybrid systems also allows for CT bone mineral densitometry. This provides useful additional information in those patients referred for bone scanning in oncological treatments that utilize hormones. The risk of osteoporosis and the potential need for bisphosphonates can be determined early as part of the staging process.

First-line investigation of trauma is usually via conventional radiography. This is generally the most available modality and is a cost-efficient first approach. When the trauma is subacute or chronic as in stress injuries, conventional radiographs may fail to demonstrate an abnormality. In these situations, the bone scan is more sensitive and can detect physiological changes before any anatomical change has occurred. This is important as many fractures or stress injuries respond well to treatment. Failure to accurately diagnose injuries can lead to complication such as pathological fracture or future degenerative change.

SPECT [61] and SPECT/CT are of proven benefit in imaging the spine with numerous studies extolling its virtue in spondylolisthesis [62]. SPECT detected 27 % more occult injuries, such as pars fractures and small joint arthropathies, over

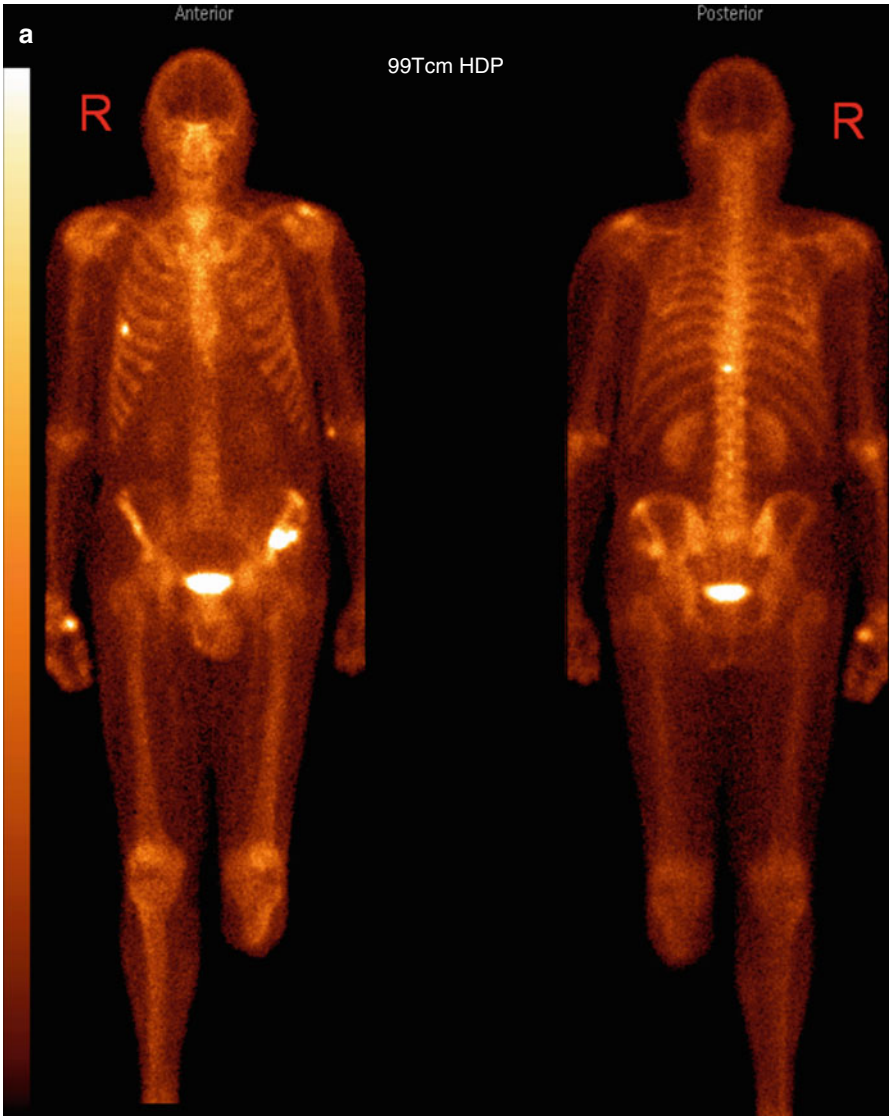


Fig. 8.26 Patient with left hip pain. (**a**) Planar images from bone scan showing increased activity in the left iliac bone, T11 vertebral body, and right 5th rib anteriorly. A left below-knee amputation is present from vascular disease. Appearances suggest bony metastatic disease. (**b**) SPECT images of the pelvis demonstrate superior localization of the abnormality to the iliac bone. (**c**) SPECT/CT shows activity localized to the iliac wing anteriorly, better demonstrated in (**d**) showing the destructive soft tissue lesion. Appearances were due to metastatic renal carcinoma, the lesions at T11, and in the rib that also represented metastases

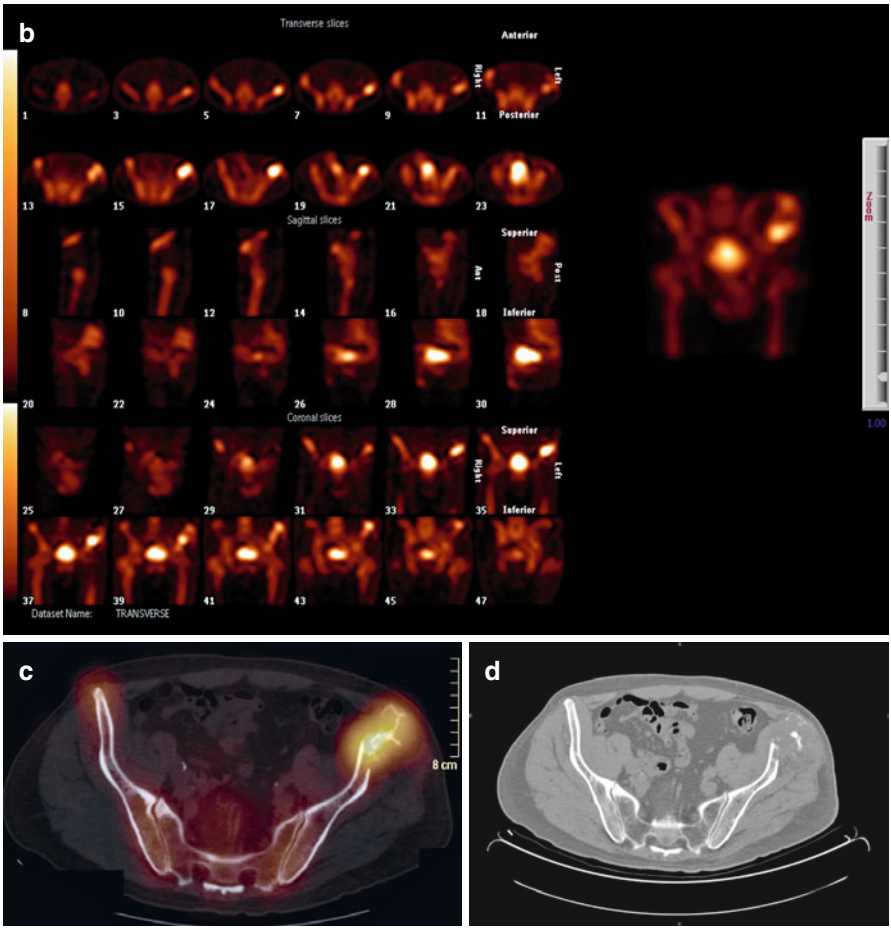


Fig. 8.26 (continued)

conventional bone scan including in some cases where the plain films were normal.

SPECT/CT is of use in investigating hip pain (see Fig. 8.27) and can demonstrate a range of conditions. Findings such as femoral acetabular impingement syndromes can be diagnosed, but MRI and MRI arthrography are found to be superior techniques for this in routine practice.

In the knee, MRI is the mainstay of imaging, but SPECT scintigraphy has been demonstrated to be able to direct an arthroscopic approach, reducing the time needed for the intervention. Adding the CT component here allows better lesion localization and delineation, identifying those lesions with loose fragments that need intervention [62]. Scintigraphy is seldom used in the shoulder due to significant overlying soft tissues whose injuries predominate and the superior ability of USS and MRI to demonstrate these injuries. Two areas that SPECT/CT offers the most impact are in the investigation of carpal fractures and in investigation of disorders of the foot and ankle.

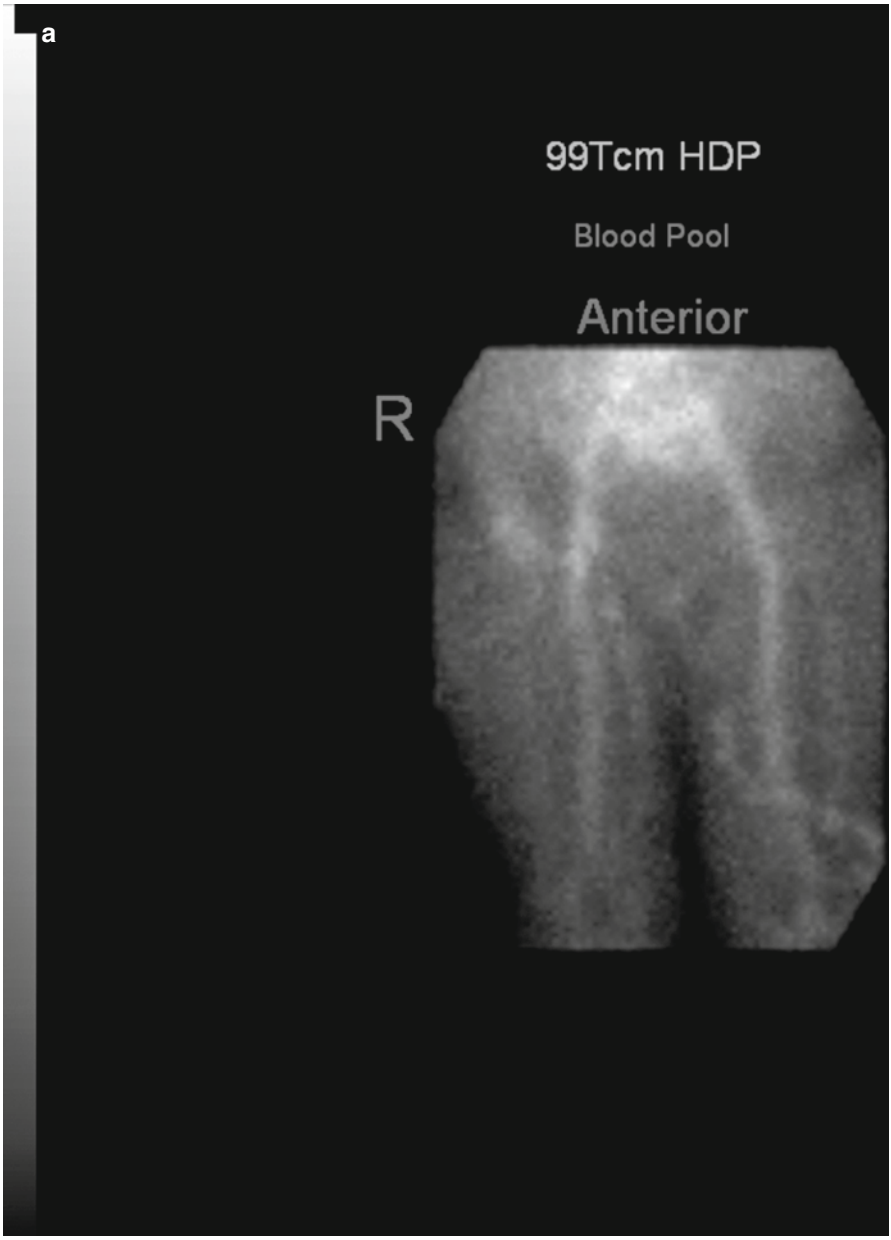


Fig. 8.27 Patient with right hip pain following a fall. Conventional radiographs were inconclusive. (**a**) Blood-pool images of the pelvis show early activity in the region of the femoral head. (**b**) Planar images show increased activity in the region of the femoral head/neck. (**c**) SPECT/CT confirms activity located to the femoral head with some disruption seen to the greater trochanteric region. This is better appreciated on the CT component (**d**). The diagnosis is of an avulsion fracture of the greater trochanter

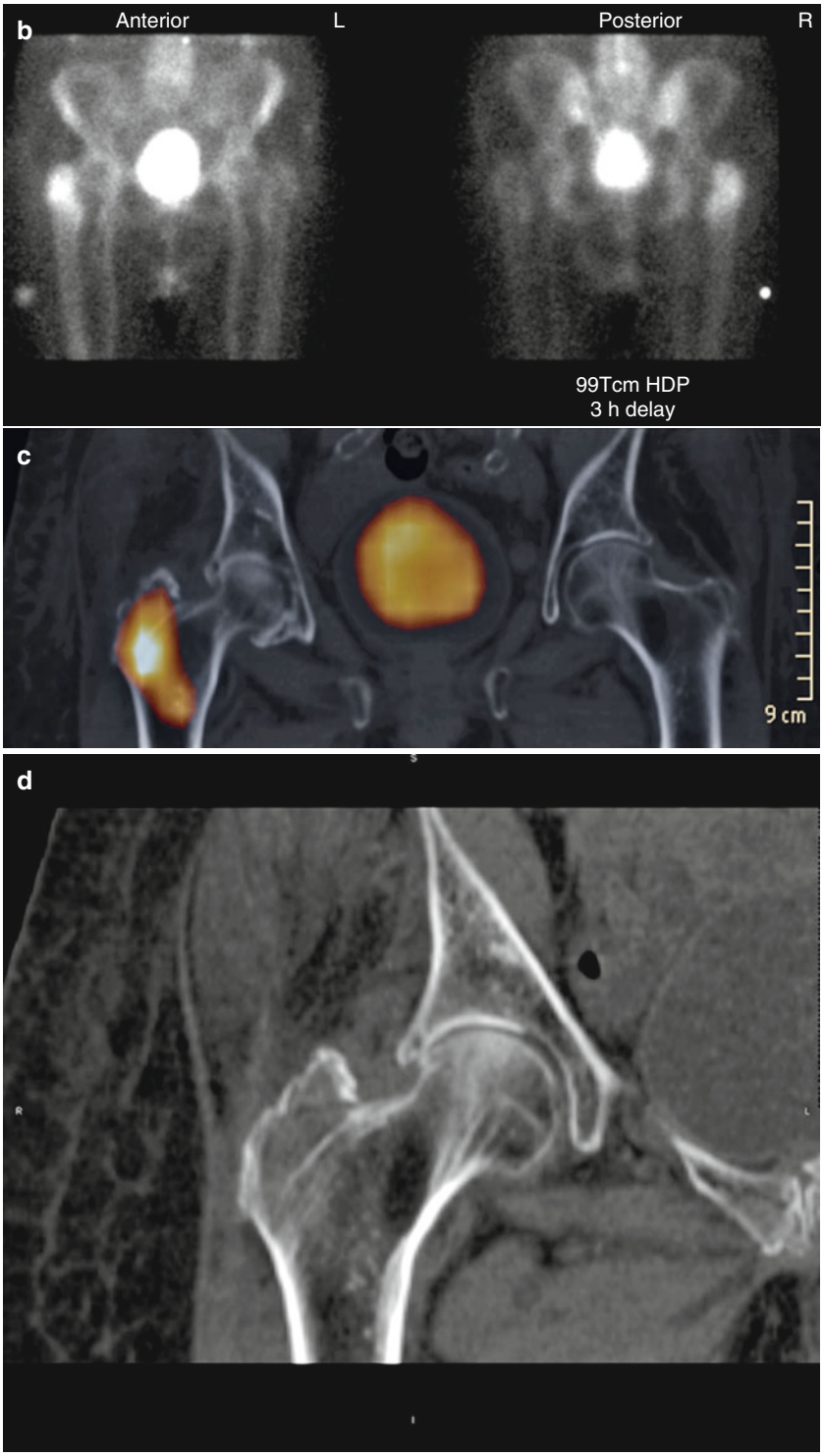


Fig. 8.27 (continued)

Use of SPECT/CT is increasing in the assessment of wrist injuries; see Fig. 8.28. Falls are common, and the wrist is frequently injured via this mechanism. The carpal bones are often involved, the scaphoid bone being the most commonly fractured carpal bone. Scaphoid fractures can be difficult to diagnose as clinical examination is often unreliable and plain films used for first-line evaluation are insensitive in undisplaced fractures. Given the impact of late complications from missed fractures of the scaphoid such as avascular necrosis, which can lead to early degeneration, this is an important area for early diagnosis.

Studies such as the one by Groves et al. [63] show scintigraphy to be more sensitive than plain films or CT alone, and the combination of scintigraphy and the anatomical information from CT is a powerful diagnostic tool. MRI is another investigative choice which has been proven to be of great value in investigating scaphoid trauma. Many centers routinely utilize the bone scan with SPECT/CT for investigating potential scaphoid/carpal injuries, but MRI is preferred in the young or where there already exists a significant degenerative change; see Fig. 8.29.

The use of SPECT/CT in the investigation of foot and ankle pathology is becoming more prevalent with a number of papers now written on this topic [64]. Indications include assessing impingement syndromes, plantar fasciitis, talar fractures and osteochondral injuries, tarsal injuries including stress fractures, and the accessory ossicles for sesamoiditis. SPECT/CT is often useful to help target the symptomatic joint in the polyarthropathic degenerate foot; see Fig. 8.30. It is used in postoperative studies to assess the success of tarsal fusion and to evaluate for postoperative complications. Authors of this chapter assessed the additional value of SPECT/CT in a small group of patients referred via a specialist clinic and found the study added additional information in 81 % of cases resulting in increased confidence and specificity in diagnosis, leading to changes in management of the underlying condition.

The increased sensitivity and localization afforded by SPECT, allied to the anatomical information afforded by CT, make for a powerful diagnostic tool. A number of difficult diagnostic conundra have been successfully resolved using this approach. Figure 8.31 demonstrates abnormal activity located at the origin of the right plantar fascia in a patient with persistent heel pain. Plain films were normal, and the patient had a contraindication to MRI by way of pacemaker. The diagnosis of plantar fasciitis was confirmed by ultrasound, and the patient was successfully treated with a guided steroid injection. This approach was particularly useful when MRI is contraindicated or unavailable and has been also been demonstrated by Breunung et al. [65].

SPECT/CT also allows for targeting actively involved joints for therapeutic injection in the polyarthropathic degenerate foot. Diagnoses of sesamoiditis and arthropathy in a congenitally bipartite cuneiform are notable successes in difficult to diagnose conditions. This is in keeping with the findings of Pagenstert et al. [66]. In their paper, 20 consecutive patients with pain of uncertain etiology were assessed by SPECT/CT. They compared the reporting confidence of a radiologist, two consultant orthopedic surgeons, and two orthopedic residents. Plain films, planar nuclear images, and SPECT/CT were all compared with all groups demonstrating improved performance and agreement with fused SPECT/CT. The fusion datasets were demonstrated to be superior to having both CT and SPECT datasets available

separately, and this was deemed of statistical significance, fusion appearing to overcome a lack of experience in the junior reporters.

Musculoskeletal infection is a common occurrence. This mostly, in uncomplicated disease, affects the skin and superficial tissues. As such, conventional radiography is insensitive for this, conventional imaging relying on demonstrating anatomical changes which are a relatively late feature of the disease process.

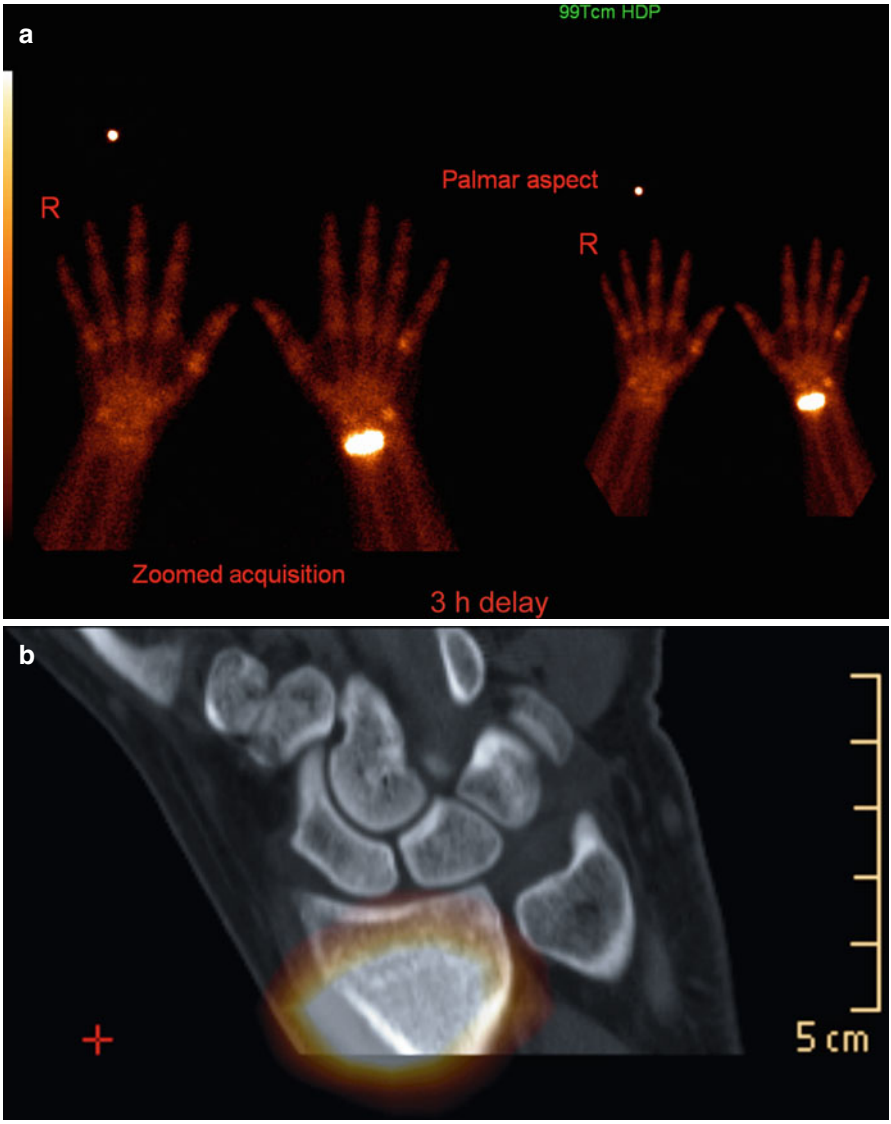


Fig. 8.28 Investigation of left wrist pain following fall onto outstretched hand. Negative conventional radiographs. (**a**) Planar images with band of increased activity in the distal radius consistent with a fracture. (**b**) SPECT/CT localizes to distal wrist although there is some slight misregistration. (**c**) CT demonstrates the fracture and proves intra-articular extension

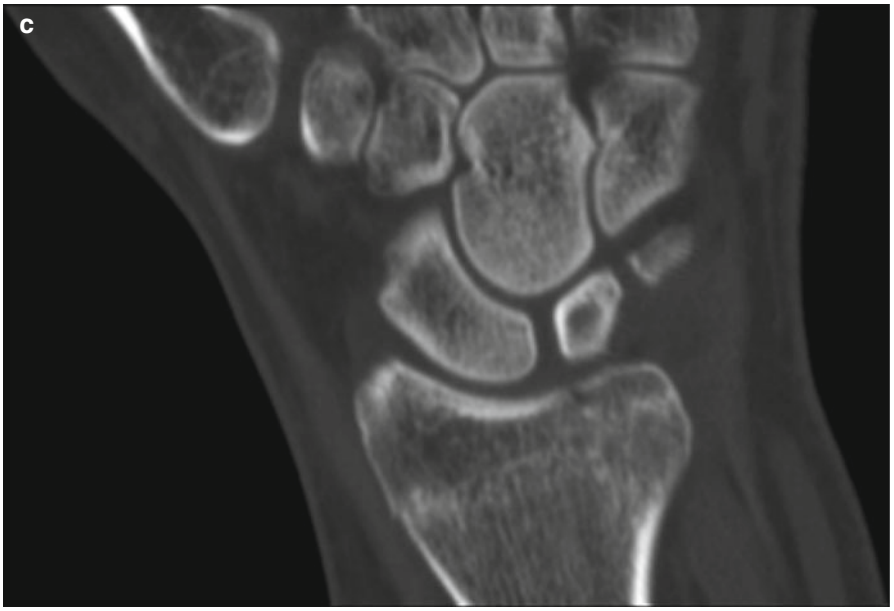


Fig. 8.28 (continued)

Functional imaging has the ability to detect earlier and therefore more treatable disease. Conventional radiographs are reserved for looking for complications such as osteomyelitis. The sensitivity of conventional radiographs is low, films often appearing normal even 14 days after the onset of symptoms. CT, USS, and MRI are superior to conventional radiographs and can be better utilized when the area of concern is defined. Often such infections have ill-defined symptoms such as PUO or persistently raised inflammatory markers, and it is difficult to target the area to be imaged. The skeletal scintigram offers the opportunity to image the whole body while performing a 3-phase examination of the affected/symptomatic area that can often demonstrate an inflammatory/infective response [67].

There are many ways to attempt to image for infection using nuclear isotope imaging [68]. The standard 99mTc-HDP bone scan is often utilized. Planar images have good sensitivity for increased osteoblastic activity but suffer from poor specificity. Overall performances are found to be in the range of 78–84 % for sensitivity with values of 33–50 % for specificity. Hybrid SPECT utilization has been shown to improve this to 86 % [69, 70].

The painful or potentially infected joint prosthesis presents a difficult diagnostic challenge to the clinician. Symptoms are often nonspecific, and plain films are often inconclusive. Conditions such as loosening or infection are not well demonstrated on conventional radiography, but this remains of some benefit in prosthetic malposition and is in routine use as the first investigation. Accurate diagnosis is crucial in determining the best treatment. This is an important consideration as we need to avoid any potentially unnecessary revision surgery.

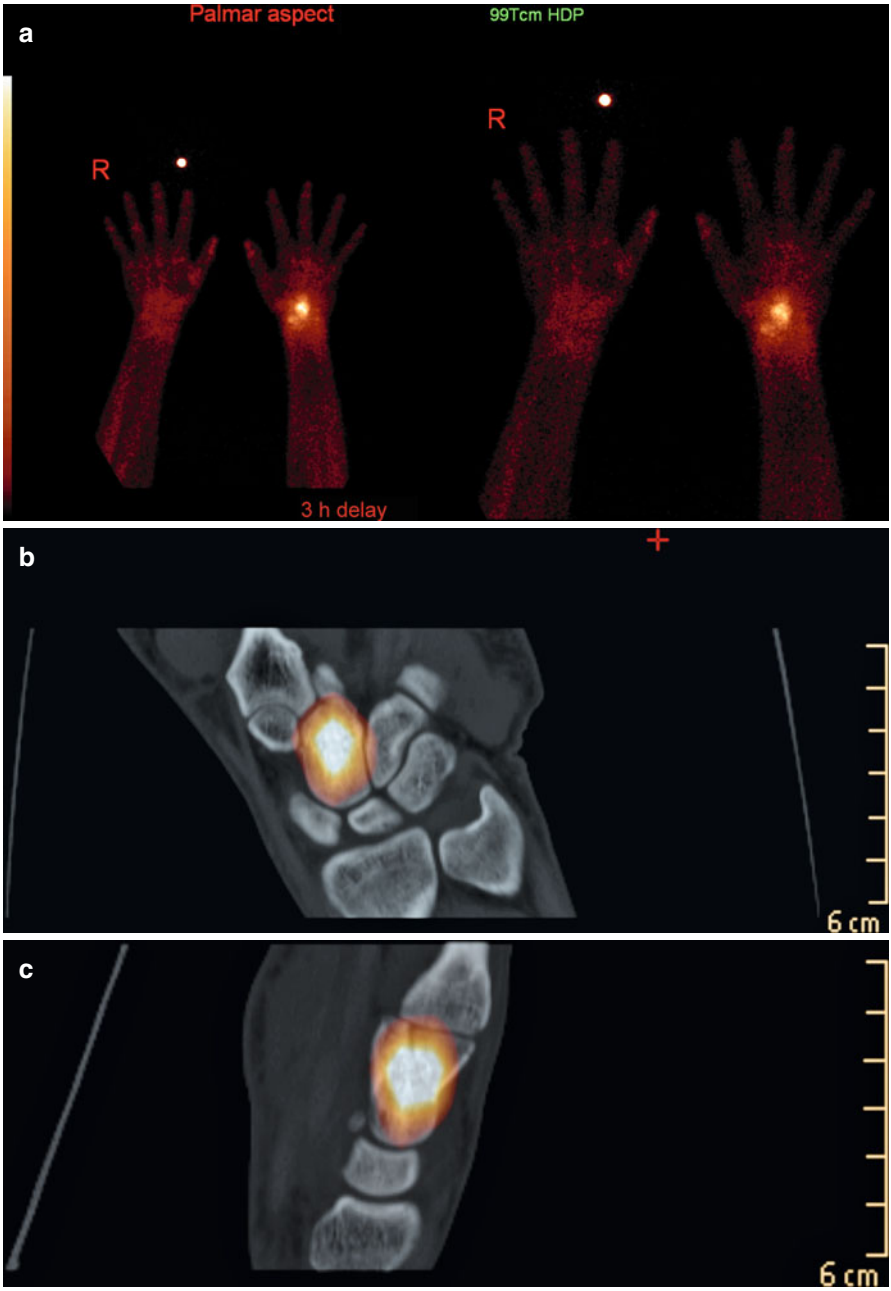


Fig. 8.29 Carpal pain following a fall. Clinical suspicion was for a scaphoid fracture, but conventional radiographs failed to demonstrate one. (**a**) Planar bone scan shows low-level uptake in the region of the scaphoid but avid uptake in the midcarpal region. (**b**) Coronal and (**c**) sagittal SPECT/CT shows the activity to be located in the capitate. (**d**) Sagittal CT component depicts a minimally displaced intra-articular fracture of the capitate

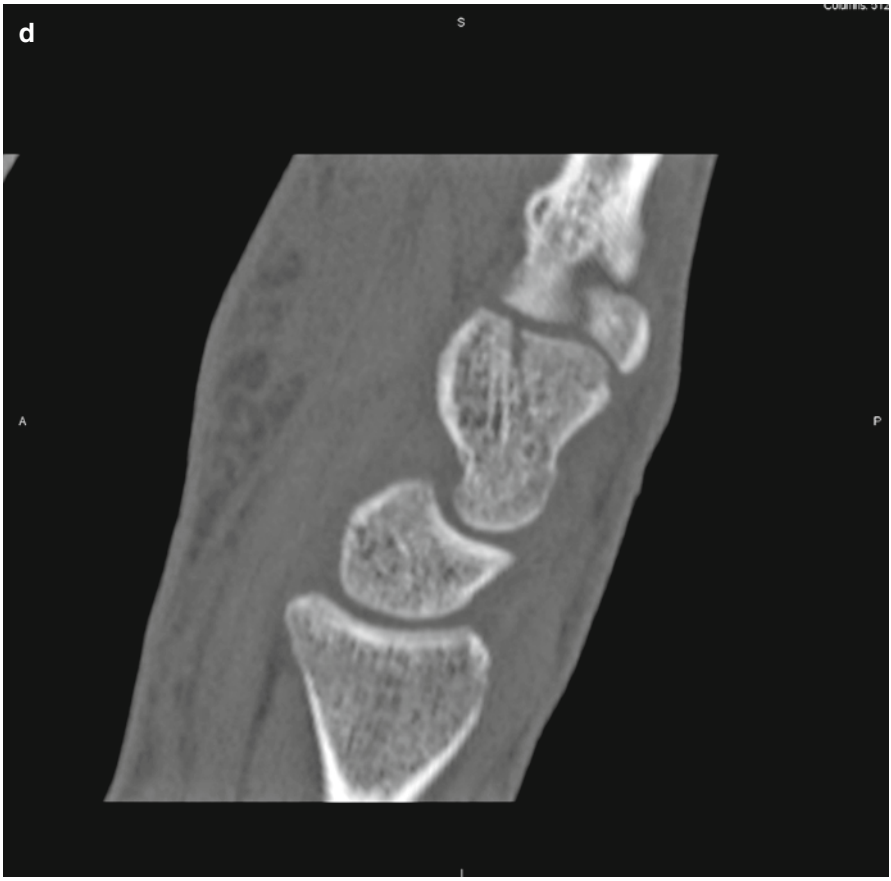


Fig. 8.29 (continued)

The ability of SPECT/CT to combine the physiological and anatomical information of both modalities has been demonstrated to be accurate in making a firm diagnosis and is in increasing use. In their paper on the value of SPECT/CT in the evaluation of patients with painful knees following total knee arthroplasty, Hirschmann et al. [71] demonstrated a change in the clinical diagnosis and therefore the proposed treatment following SPECT/CT in 19 out of 23 cases. Conditions such as loosening and progressive osteoarthritis were demonstrated along with periprosthetic fractures, synovitis, and component malrotation.

Gallium-67 citrate and 111In-labeled leukocytes are alternative tracer agents that can also be used to demonstrate areas of inflammation/infection. Gallium-67 citrate is useful in excluding an inflammatory or infective process but is less often utilized due to suboptimal imaging characteristics including increased dose. The need to employ delayed imaging is also less acceptable to the patient. The ability of scintigraphy to demonstrate complications of the infective/inflammatory process is limited

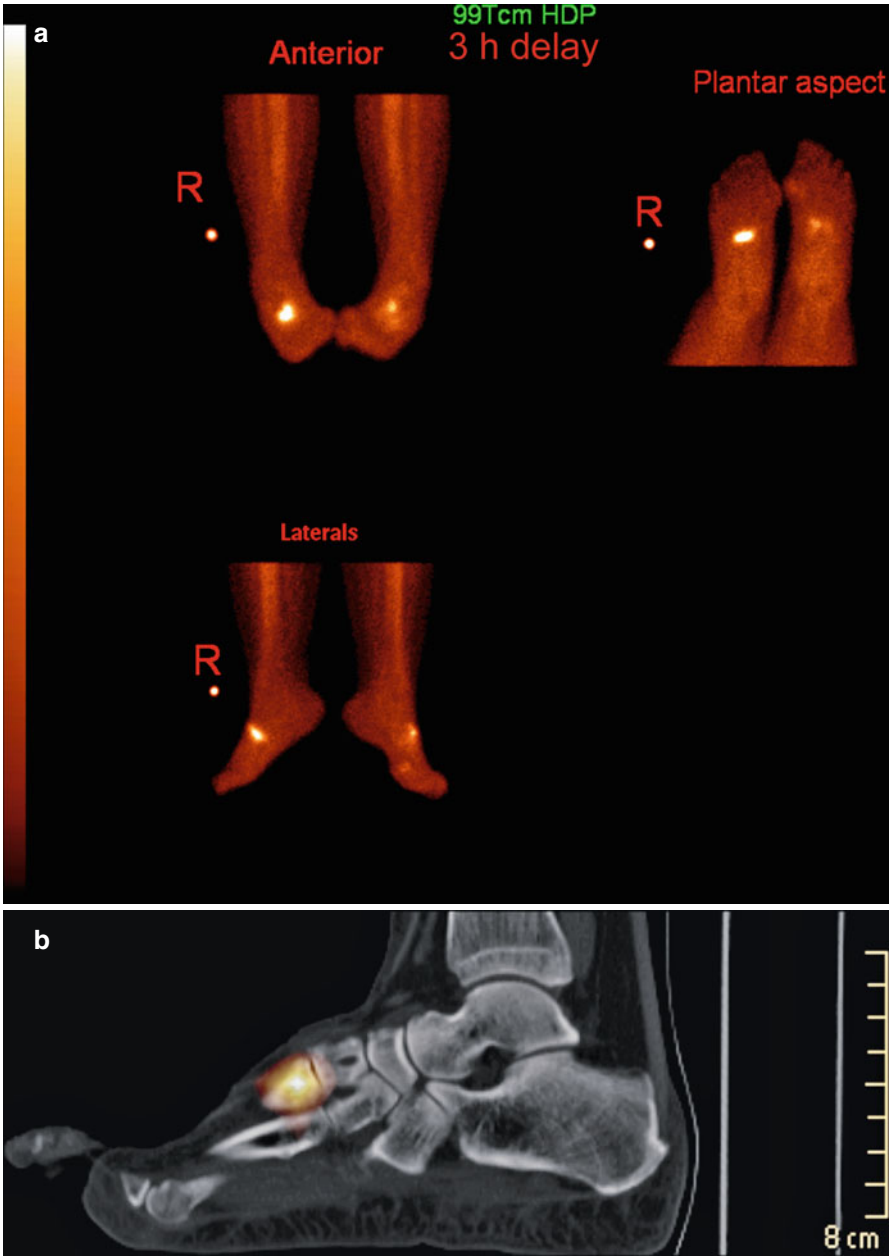
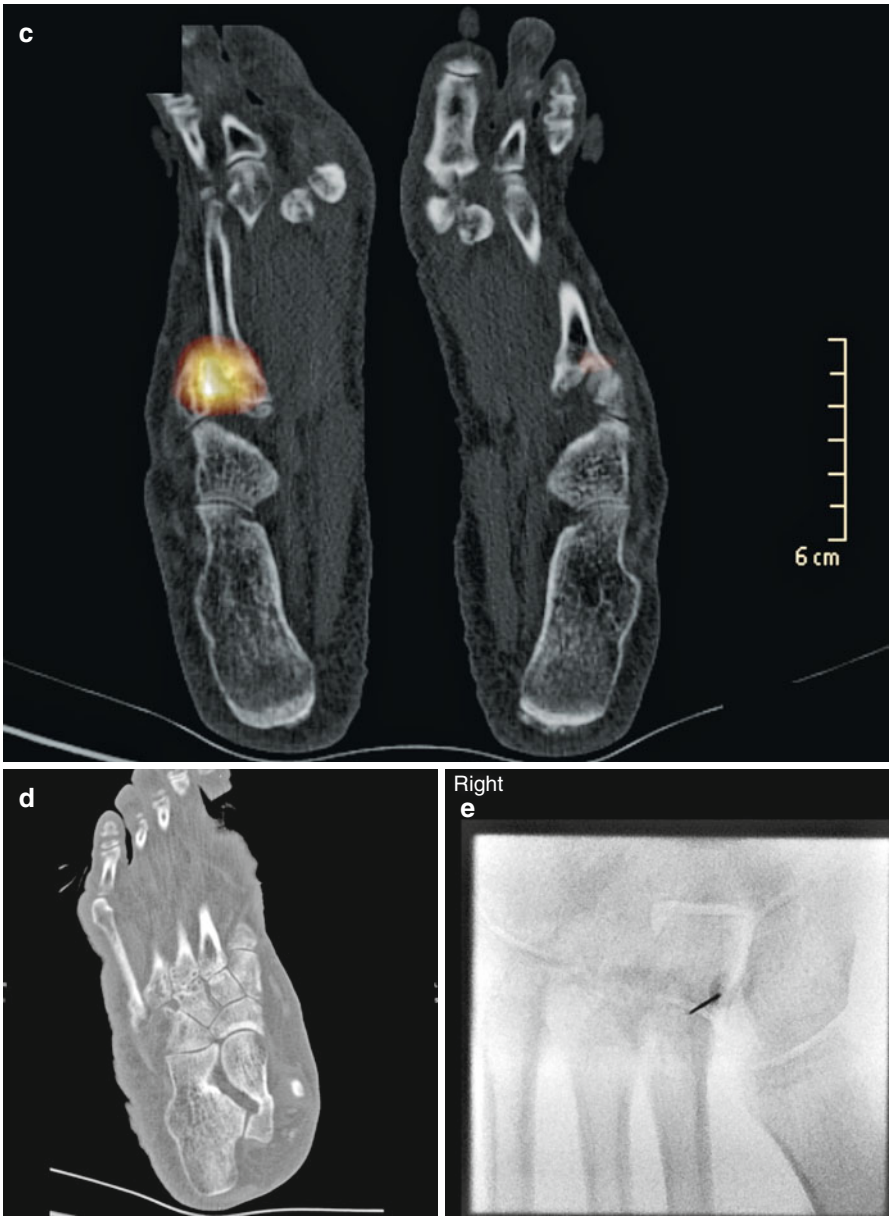


Fig. 8.30 Polyathropathic foot with worsening right midfoot pain. Bone scan was utilized to attempt to identify which joint was likely to be symptomatic. (**a**) Planar bone scan shows increased activity in both midfoot regions but particularly on the right midfoot. (**b**) Sagittal and (**c**) coronal SPECT/CT shows activity to be located at the cuboid/cuneiform metatarsal joints. (**d**) Coronal CT demonstrates degenerative change in this area. (**e**) Fluoroscopic-guided joint injection of this area successfully alleviated the patients pain



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by poor anatomical resolution. It is here that SPECT/CT has an increasing role, particularly in assessing spinal involvement [72].

Radiolabeled leukocytes is a sensitive method with 111In-labeled leukocytes previously in widespread use [73]. ^{99mTc}-HMPAO and 99^{m}Tc-labeled leukocytes

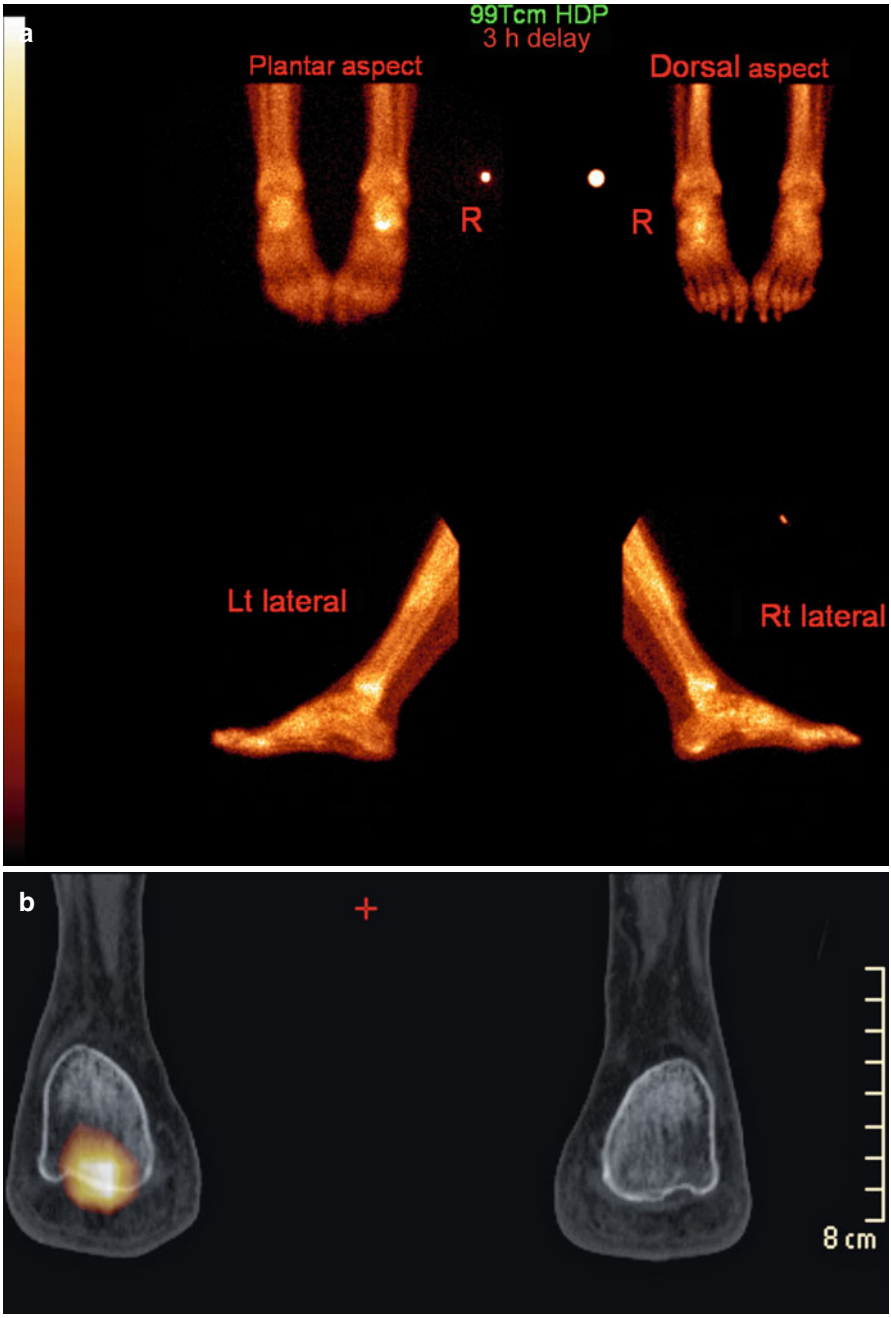
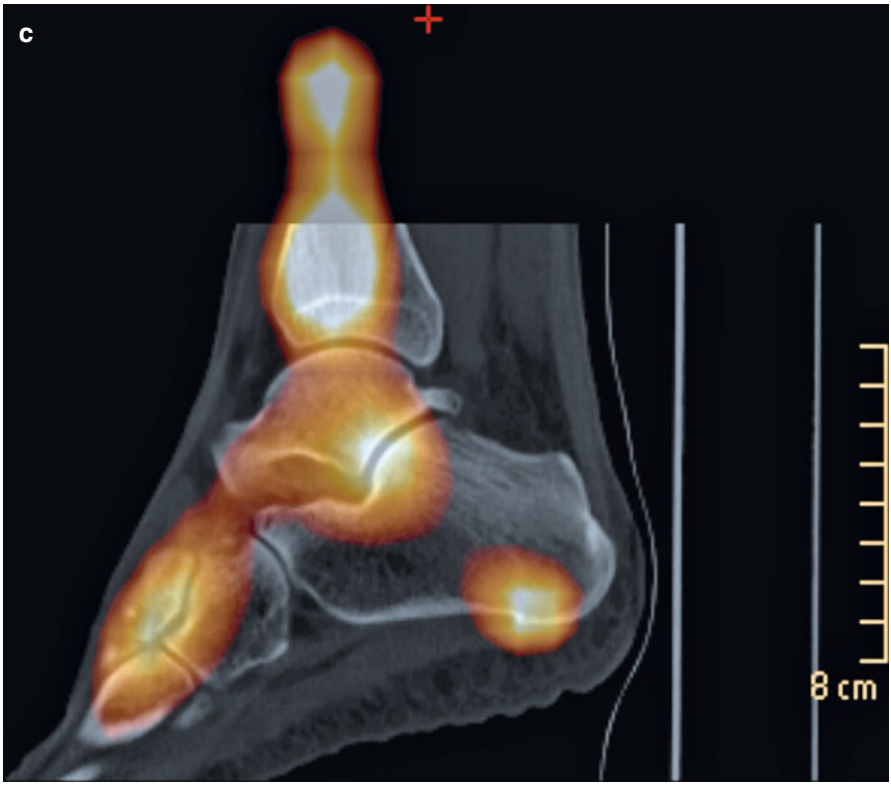


Fig. 8.31 Right heel pain. (**a**) Planar images show subtle increased activity on the right at the origin of the plantar fascia. (**b**) Coronal SPECT/CT confirms the activity on the right, inferior to the calcaneum. (**c**) Sagittal SPECT/CT of the right foot demonstrates activity at the origin of the plantar fascia. The patient had successful treatment for plantar fasciitis



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have been developed with superior imaging characteristics and are increasingly utilized. The impact on radiopharmacies however with complex laboratory preparation remains an issue, particularly in the smaller imaging departments.

Combinations of isotopes are seen to give the best results with 111In-leukocytes coupled with ^{99m}Tc-HDP being good for osteomyelitis and 111In-leukocytes coupled with ^{99m}Tc-sulfur colloid of proven benefit in bone and joint infections [68]. The added value of SPECT/CT in improving specificity is well documented [69, 70, 74].

The use of these isotopes is not limited to investigating musculoskeletal infections and is useful in the investigation of pyrexia of unknown origin (PUO) [75] and in the assessment of possible vascular graft infections [72, 76]. SPECT/CT improves localization and reporter confidence in these areas as has been demonstrated for MSK infection.

SPECT/CT is also useful in the investigation of malignant otitis externa. Bony involvement of the inflammatory/infective process is important to determine as the treatment options vary with the extent of disease. The CT component usefully quantifies any bony destruction and can be used to look for intracranial complications. This approach is increasingly utilized and reported [77] (see Fig. 8.32).

Newer tracer agents offer promise for further gains in specificity [78], while PET/CT shows much promise but needs further validation.

Metabolic diseases of bone often have biochemical and morphological changes which can be seen on plain films and are the mainstay of diagnosis. Osteoporosis is characterized by reduction in bony density with bone scanning not routinely utilized. Attempts at quantitative bone scans and quantitative SPECT have demonstrated disappointing results and are not routinely used. The CT component of the hybrid scanners does however allow for quantitative CT bone mineral densitometry measurement which we have found useful in assessing the osteoporosis risk in breast cancer prior to hormonal therapy.

Similarly, there is no routine role for bone scanning in osteomalacia, but it can be used in the assessment of complications of this condition such as pathological fractures; see Fig. 8.33.

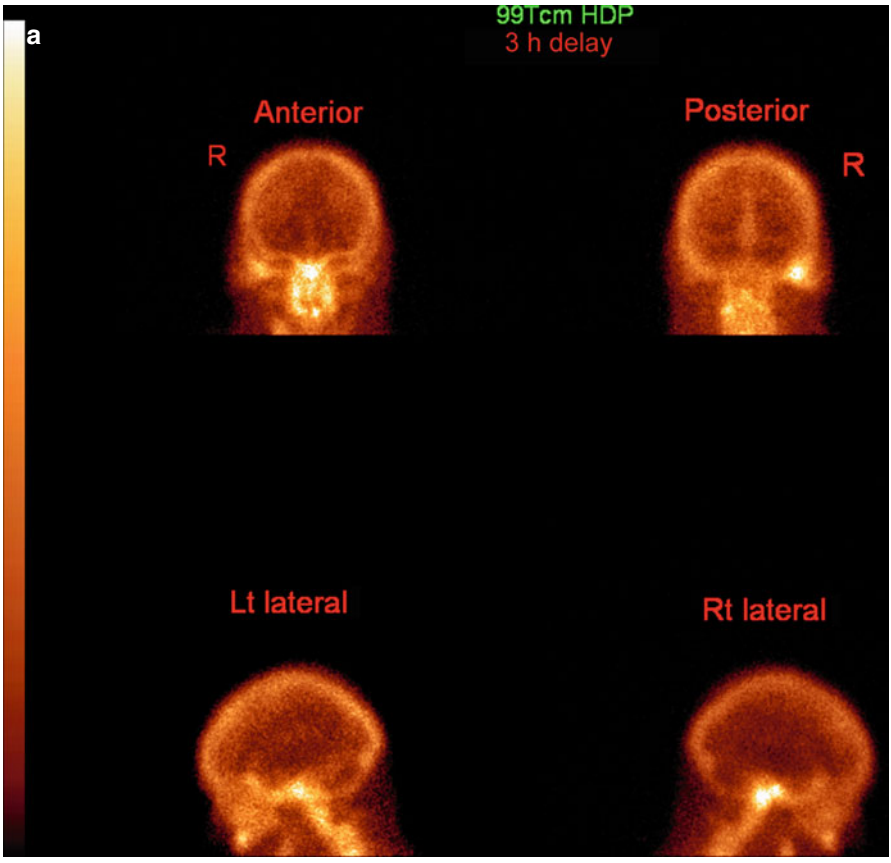
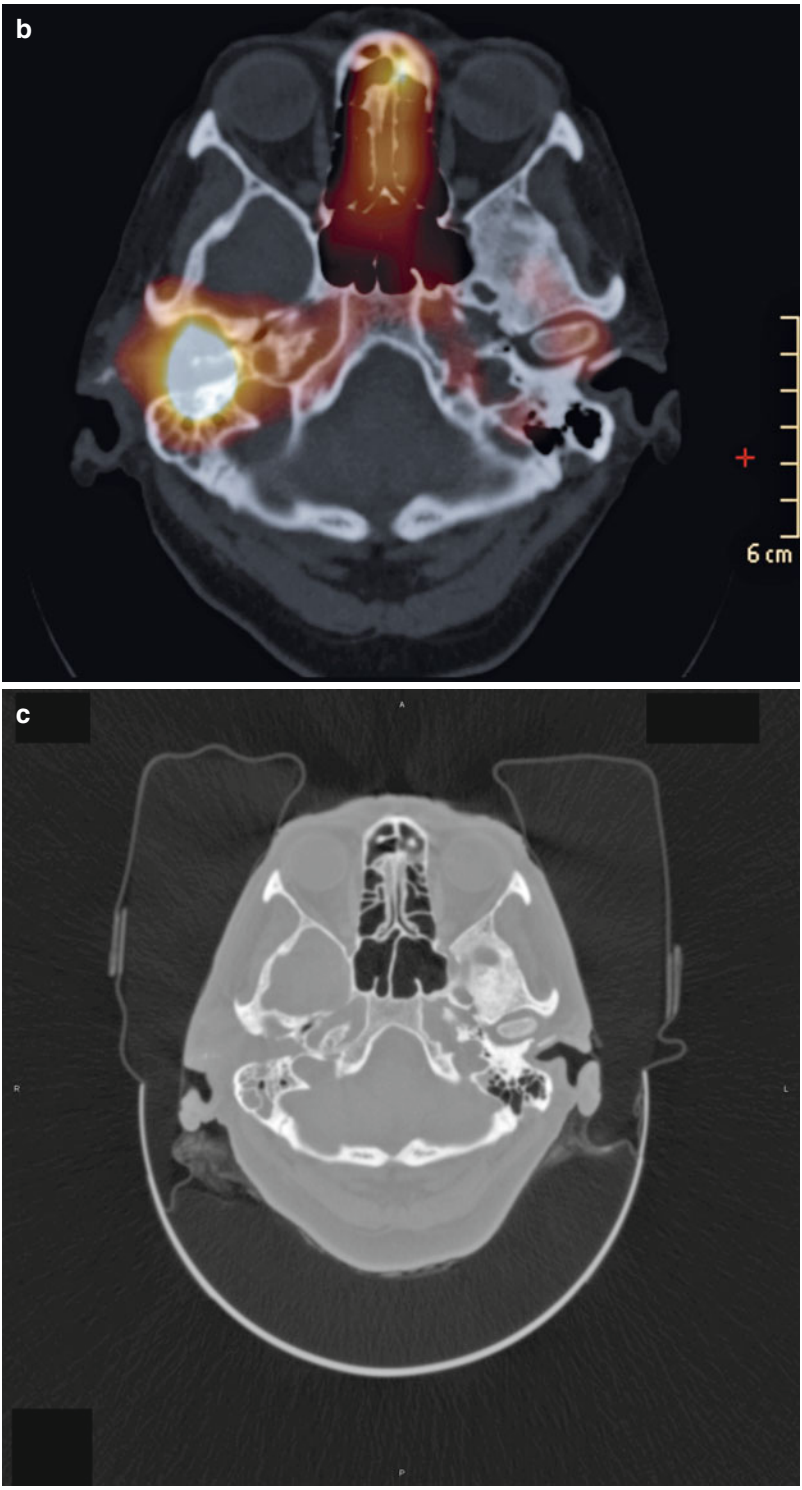


Fig. 8.32 Malignant otitis externa. Patient with painful right ear. Bony involvement was suspected clinically. (**a**) Planar images demonstrate increased activity in the region of the right mastoid. (**b**) SPECT/CT and (**c**) CT images show abnormality to be localized to the external ear canal and mastoid region with some clouding of the mastoid air cells and bony erosion consistent with malignant otitis externa



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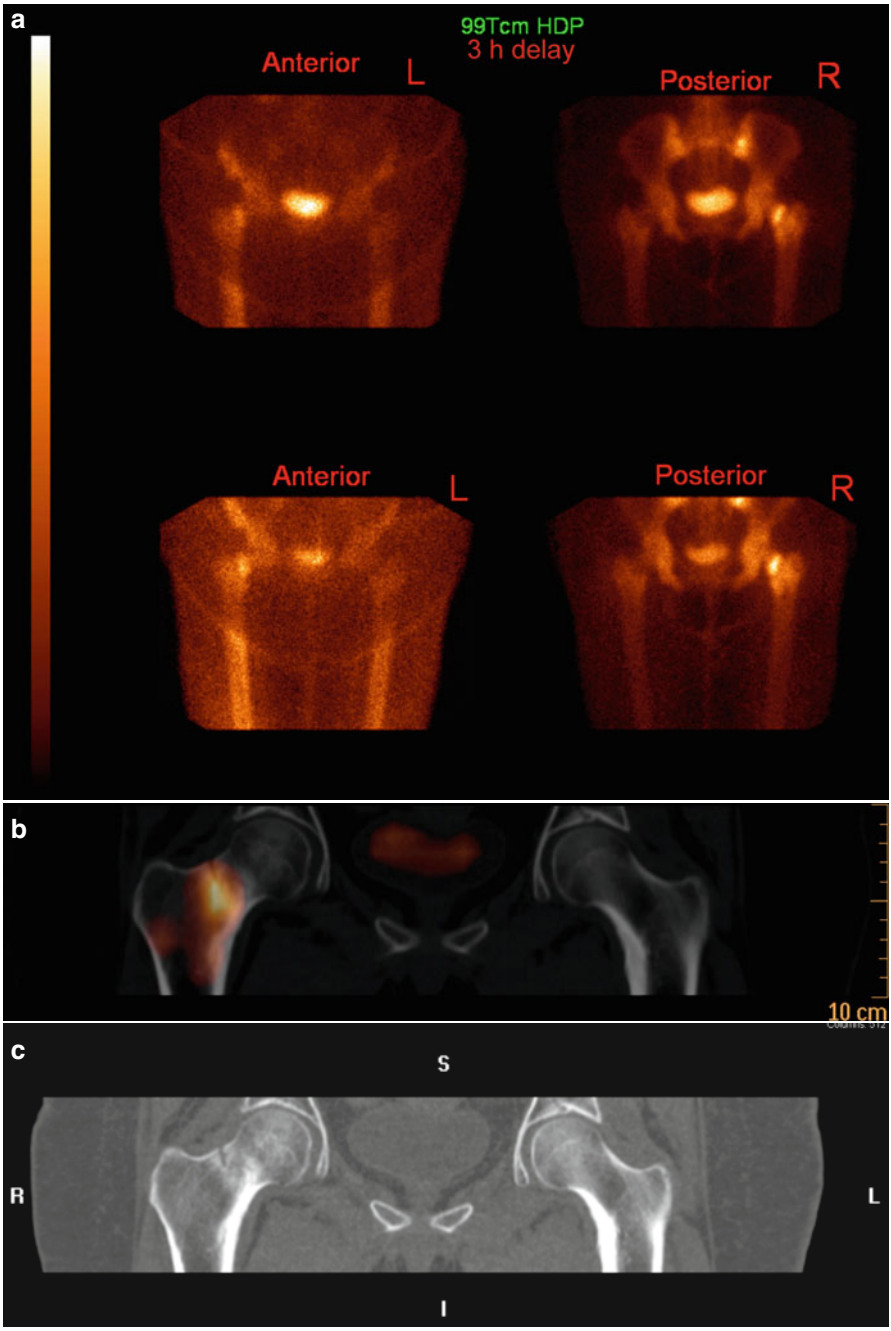


Fig. 8.33 Stress fracture. Patient with long-standing hip pain. (**a**) Planar images showing increased activity in the right femoral neck. (**b**) SPECT/CT and (**c**) CT showing incomplete stress fracture through the right femoral neck

The bone scan gives a good overview of the distribution of arthropathy, both degenerate and rheumatological; see Fig. 8.34. SPECT/CT being reserved for those areas of diagnostic uncertainty such as the spine, pelvis, and wrists/hand and ankle/feet.

SPECT/CT is not a routine investigation in Paget's disease of bone as the planar imaging pattern is usually sufficient when combined with plain films for correlation. For areas of diagnostic doubt such as vertebral involvement, the CT component can be useful.

Newer treatments such as intravenous bisphosphonate therapy are being used to treat Paget's disease. Bone scanning is useful to demonstrate the extent of bony involvement and can be usefully used to demonstrate any response to therapy. SPECT/CT has no routine role in this but can be useful if a complication of Paget's such as pathological fracture or malignant change is suspected [79, 80].

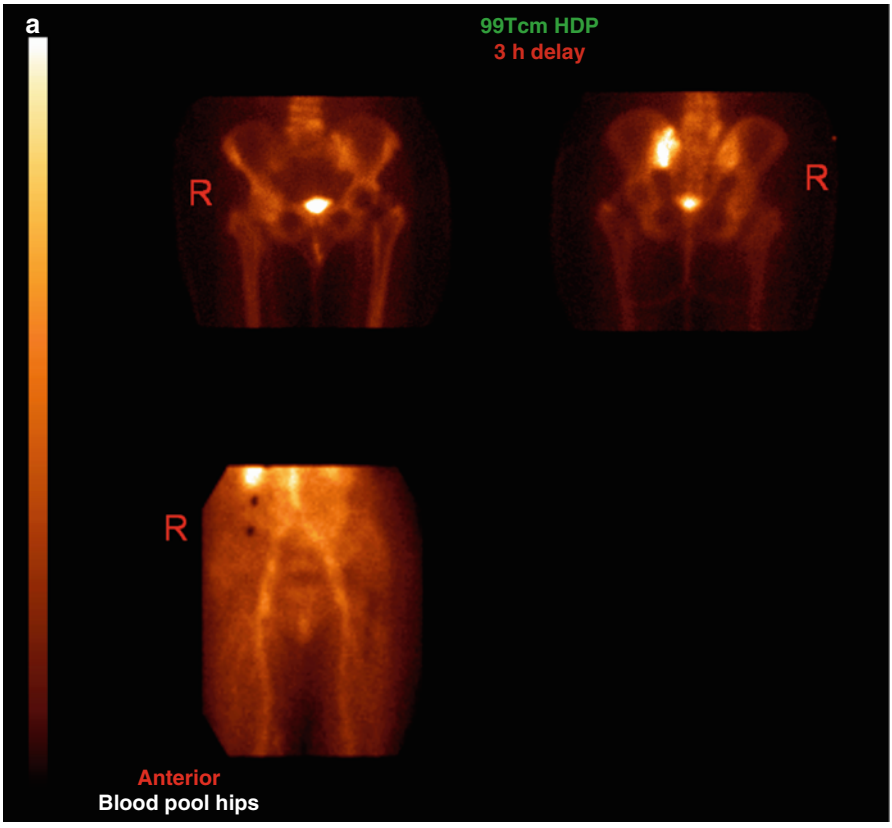
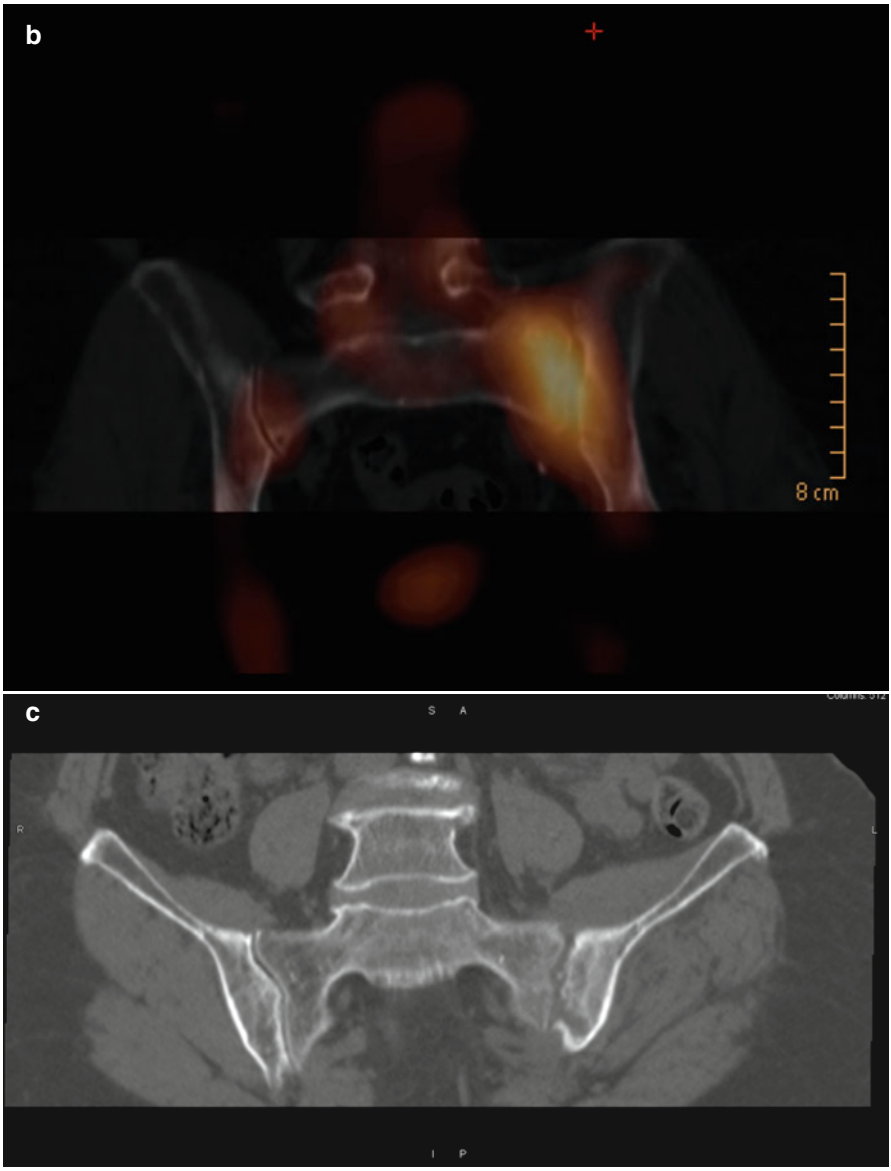


Fig. 8.34 Sacroiliitis. (**a**) Planar images demonstrating increased activity in the region of the left sacroiliac joint. (**b**) SPECT/CT confirms sacroiliac joint pathology, and (**c**) CT component showing increased joint space and bony erosion in keeping with a sacroiliitis



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Skeletal scintigraphy remains a sensitive examination in children, but the lack of specificity and the presence of avid growth plates restrict its routine use. Hybrid imaging offers the opportunity to overcome this lack of specificity but comes at a dose penalty. Examinations in children are more often tailored to each patient with direct supervision of the study to optimize dose rather than

imaging to a set acquisition protocol. Imaging in children is also more problematic with movement during the longer nuclear acquisitions a common occurrence. Movement can lead to misregistration artifacts when images are fused. The use of sedation or even performing the studies under general anesthetic can be considered if excessive movement is likely to degrade the images too much [81].

Imaging of primary bone malignancies in children is often best performed as part of a multidisciplinary approach within specialized tertiary centers [82]. The planar images are useful to define monostotic versus polyostotic bony involvement, and SPECT is useful to clarify indeterminate lesions, particularly in the spine and pelvis. The ability to stage with CT at the same visit is of benefit, and developing one-stop staging investigations can be considered. Benign bony lesions are seldom imaged using nuclear techniques, MRI being preferred, but it can be useful to determine the extent of skeletal involvement.

Similarly, MRI is the modality of choice in imaging skeletal infection, but the bone scan can be of use in determining the distribution of the disease. This is particularly so for infants below the age of 2 where the pattern of disease dissemination is hematogenous and polyostotic disease is more frequent.

Skeletal scintigraphy is considered complementary to a plain-film skeletal survey in non-accidental injury [83]. Abnormalities on the bone scan should be correlated with further anatomical imaging. Generally, the plain radiograph will suffice, but SPECT/CT is of use in the hands and feet and also in flat bones such as the scapula and pelvis. SPECT studies are good for identifying rib fractures [81]. Correlation of abnormal bone scan findings with conventional radiographic or CT evidence of injury increases diagnostic certainty which is extremely important in this area.

The bone scan is useful in trauma such as the limping child, particularly when they have yet to develop speech. Toddler's fractures can occur anywhere from hip to foot and often present with normal initial radiographs. The bone scan will often identify the affected area, and plain-film imaging can be focused here. In areas that plain film is less sensitive such as the acetabulum or foot, SPECT/CT can be useful.

Skeletal scintigraphy is of use in assessing the adolescent with back pain. SPECT/CT is of proven use in this area with plain films often being insensitive and the bone scan and even SPECT being less specific. It is particularly useful in diagnosing and delineating spondylolysis but is of use in assessing variant anatomy which can lead to abnormal mechanical stresses and therefore be symptomatic.

Another area that SPECT/CT is seen to be useful is in assessing mandibular condylar asymmetry. Surgical correction is not contemplated while there is active growth and possibility for natural remodeling. Planar imaging can demonstrate ongoing osteoblastic activity with SPECT and SPECT/CT in particular accurately localizing this to the mandibular condyle. An added feature of HP-CT systems is the ability to generate 3D models of the mandible which can then be used to more accurately plan surgical or orthodontic approaches; see Fig. 8.35.

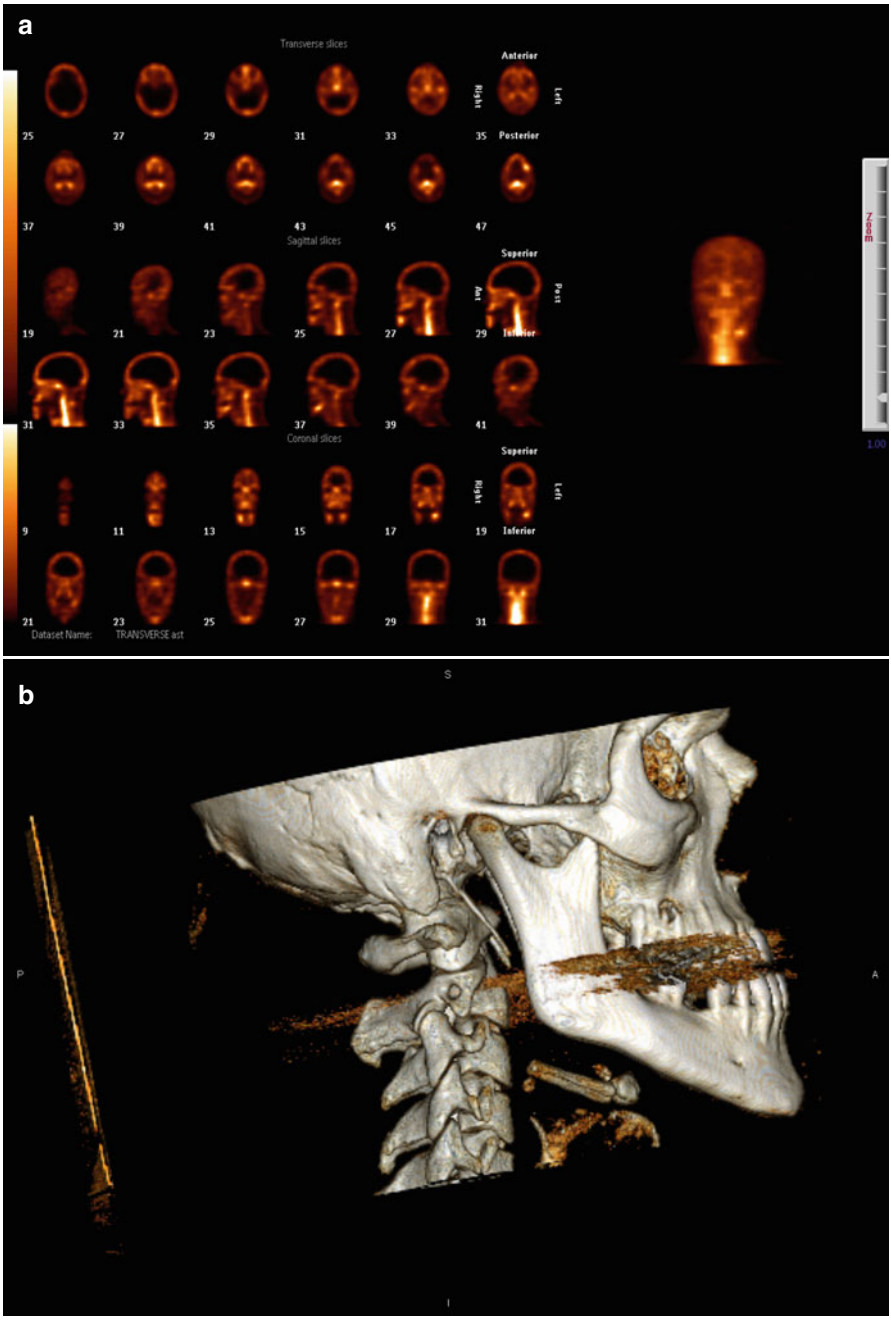

Fig. 8.35 Assessment of mandibular asymmetry. (**a**)SPECT study demonstrating ongoing bony turnover in the left mandibular condyle. This suggests ongoing potential for remodeling, and surgical correction is usually deferred. (**b**) 3D CT reconstruction from the CT dataset allows for surgical planning

8.14 Miscellaneous Uses of SPECT/CT: SPECT/CT in Lung Disease

Planar V/Q scanning has been utilized for some time in the investigation of pulmonary embolism. Since the early 1990s, however, the CT pulmonary angiogram has become the favored modality for the investigation of suspected PE [84]. The advantages of CTPA include its availability and ability to accurately diagnose many of the differential diagnoses involved. Despite its wide acceptance as an imaging test, there are a number of drawbacks. Radiation dose is higher than for nuclear techniques. The sensitivity reported in studies such as PIOPED [85] is in the region of 85 % although this is improved on the newer CT scanners available since the study was published. The test performs more poorly in pregnant patients owing to altered physiological state and flow dynamics depositing unopacified subdiaphragmatic blood into the pulmonary arterial system, diluting the administered contrast. CTPA is contraindicated in those with iodine allergy and is relatively contraindicated in renal impairment/failure.

SPECT V/Q scanning improves sensitivity and specificity over planar V/Q and is of lower radiation dose than CTPA. SPECT V/Q can reduce the amount of indeterminate results to less than 5 % and does not show incidental findings such as pulmonary nodules that require further follow-up. If a CT component is added as in SPECT/CT V/Q, there is demonstration of incidental findings, but the increase in specificity adds further to the value of the examination. The improved performance of SPECT V/Q over planar V/Q has resulted in the EANM advising this as the method of choice where possible [86, 87]; see Fig. 8.36.

The use of CT with SPECT V/Q has been varied. Some institutions fuse the isotope data to a CTPA study, whereas others fuse with low-dose CT. The principal purpose here is to add anatomical information to the nuclear study, accounting for deficits by anatomical abnormalities. This allows for changes in reports from intermediate probability to low probability with increased certainty [88].

A further use of hybrid imaging is in the scenario of a normal study. The HP-CT scanners allow for immediate HRCT in the investigation of symptoms such as shortness of breath, and this has been utilized in diagnosing hypersensitivity pneumonitis in a young female with a normal perfusion scan; see Fig. 8.37.

8.15 The Importance of Image Evaluation of the Raw CT Data

CT has undoubtedly revived interest in many nuclear techniques, being used for attenuation correction, localization, and lesion characterization. It is felt by this author that it is important to evaluate any CT component as an imaging examination in its own right, particularly so where the CT has been obtained for attenuation correction purposes. Our experience of assessing the CTAC dataset in myocardial perfusion studies has shown a number of incidental findings. Many are of no consequence, but we have found an incidence of approximately 1 % undiagnosed

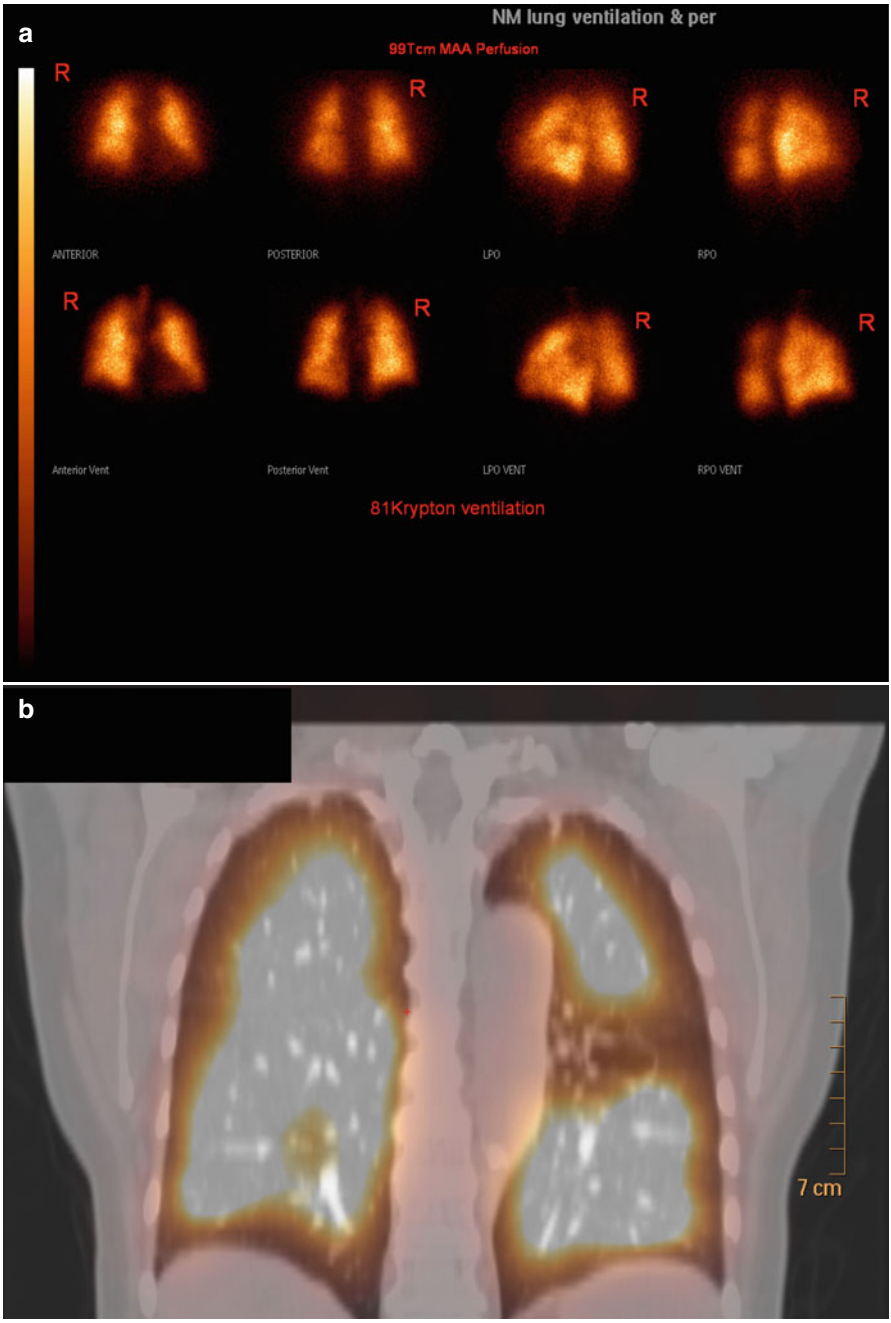

Fig. 8.36 Suspected pulmonary embolism. (**a**) Planar V/Q demonstrates a left mid-lung deficit that needs to be accounted for. (**b**) Coronal, (**c**) sagittal, and (**d**) axial SPECT/CT demonstrate defect to be due to a large pneumatocele, allowing the scan to be interpreted as normal. (**e**) axial CT images demonstrating the pneumatocele

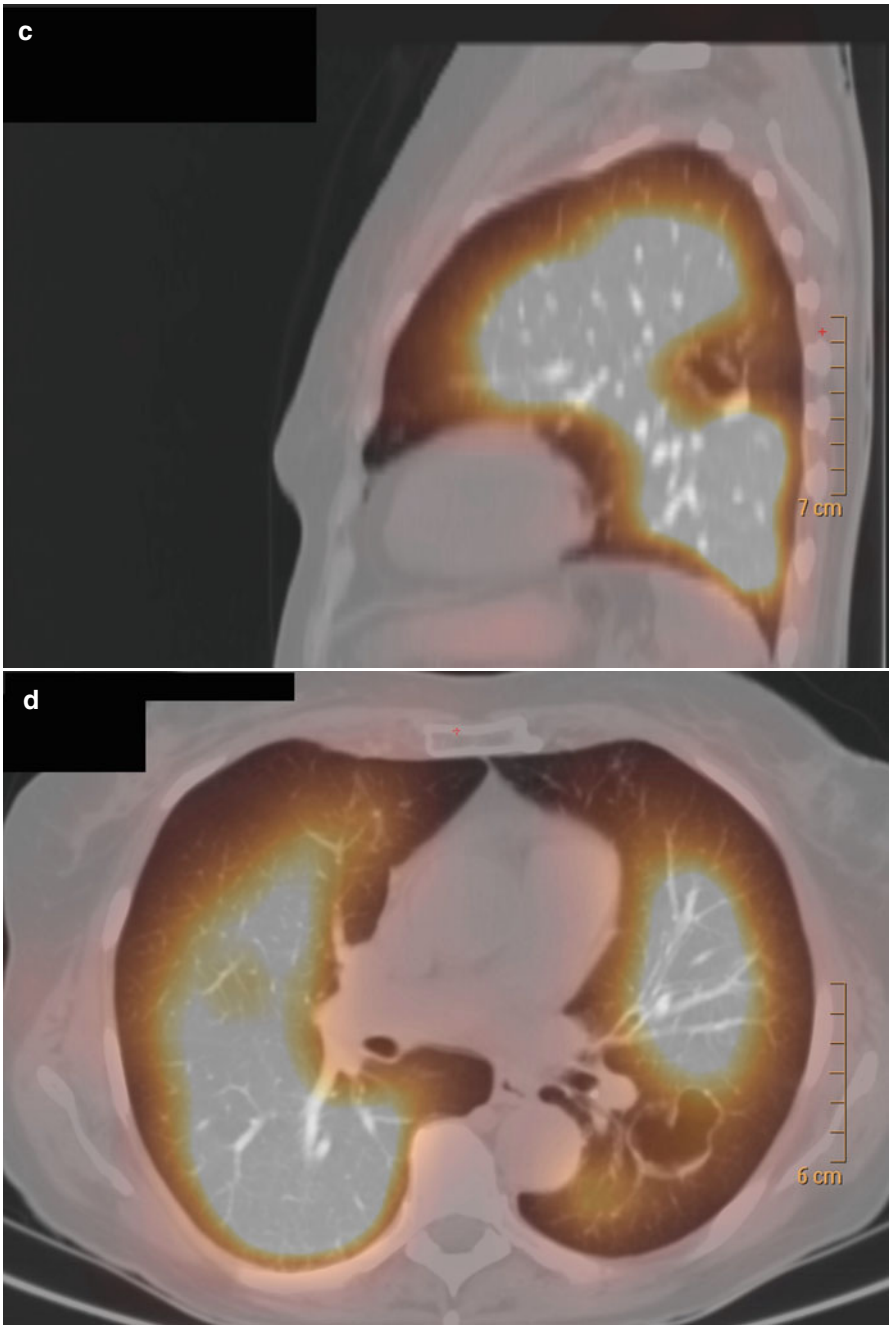

Fig. 8.36 (continued)

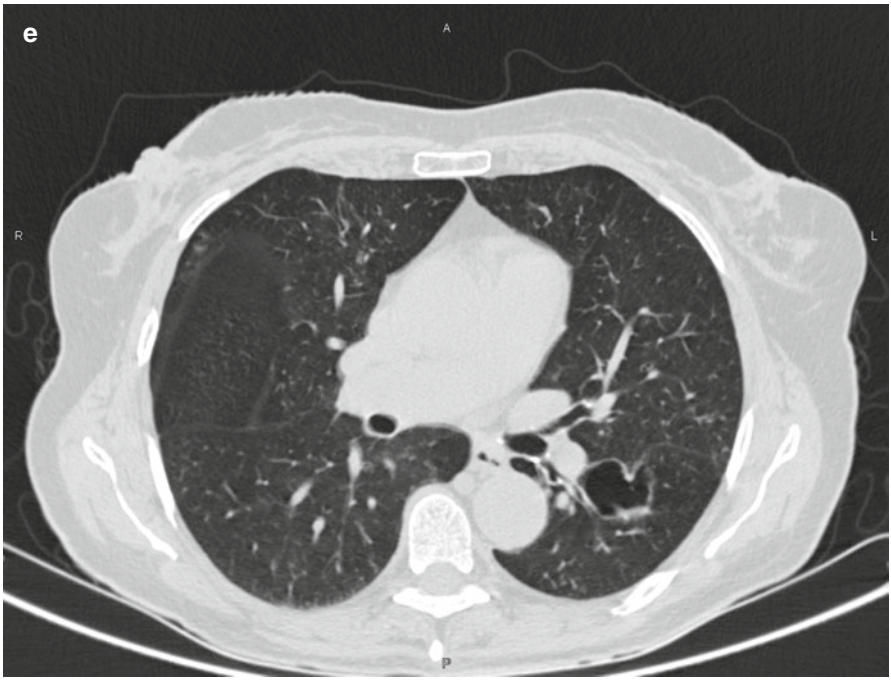

Fig. 8.36 (continued)

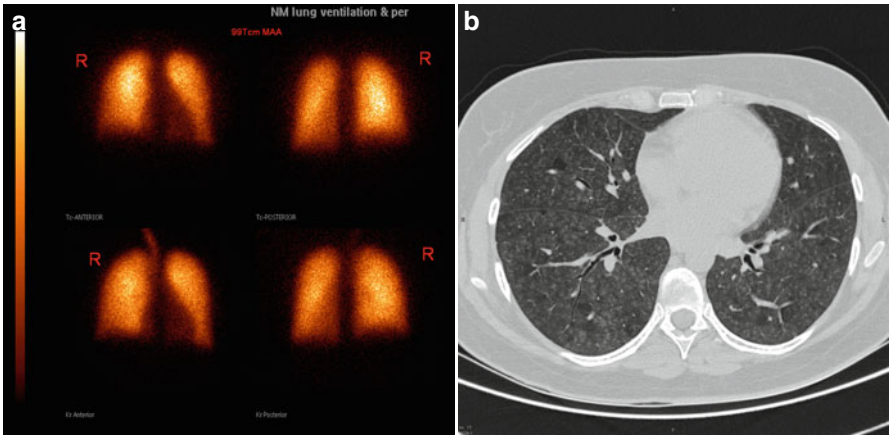

Fig. 8.37 Investigation of breathlessness. As part of the investigations for this patient, a V/Q scan was obtained to exclude PE. (**a**) The planar images are normal. The patient was very breathless, however, and HRCT was performed at the same time to further investigate. (**b**) Axial CT images demonstrate widespread centrilobular nodules in keeping with the diagnosis of hypersensitivity pneumonitis

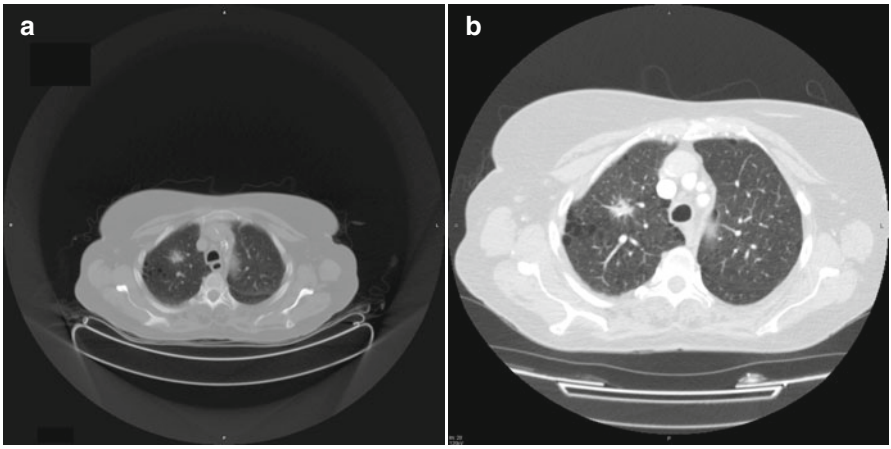

Fig. 8.38 CT attenuation correction dataset findings. (**a**) CTAC dataset showing irregular nodule in the right upper lobe. (**b**) Corresponding image from subsequent staging CT scan confirming the nodule in the right upper lobe. The eventual diagnosis was of a T1b right upper lobe carcinoma

lung cancers; see Fig. 8.38. This incidence is in line with that found in the lung cancer screening data, underlining the potential importance of image review. More work is needed in this area to assess the varying contributions of different specification systems and lesion detectability.

8.16 Summary

In this chapter, we have attempted to show how SPECT/CT can influence clinical interpretation of nuclear medicine images; there is increasing published evidence that in appropriately selected cases, the addition of hybrid imaging adds significantly to the clinical yield of the study, either by modifying the image as in cardiac attenuation correction or by adding anatomical information into the image set to improve both localization and characterization. There is still a need for further research to define the appropriate use of this technique.

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Part IV
Radiation Protection

Chapter 9
Practical Radiation Dose and Practical Radiation Protection Considerations

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Radiation dose in CT has received increased attention over the past few years, since CT delivers one of the highest doses to patients compared to other imaging modalities [1–3]. For this reason, it will be reviewed here. Only major factors will be considered.

An important first consideration is the beam geometry. As described in Chap. 5, CT uses a fan-shaped X-ray beam directed to an array of detectors that rotate 360° around the patient to collect attenuation data. The patient imaging table moves during the scanning process, and the X-ray tube traces a spiral or helical beam path around the patient. Ideally the radiation intensity measured along the z-axis would have equal intensity everywhere inside the beam and would have no intensity on either side, and it is clear that the dose distribution is almost always wider than the nominal slice width (SW). An important consideration in discussing CT dose is the dose distribution. Seeram [4] points out that the dose distribution is given by the function $D(z)$, which describes an arbitrarily shaped dose intensity along the patient axis. In general, the shape varies between CT scanners. $D(z)$ is very important to dose in CT, since it is this dose distribution that is measured.

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9.1 CT Dose Descriptors

To describe the dose in CT, several dose descriptors are used; however, only two will be described here since it is beyond the scope of this chapter to address all aspects of CT radiation dose. These include the volume CTDI ($CTDI_{vol}$) and the dose length product (DLP).

9.1.1 Computed Tomography Dose Index (CTDI)

The first definition of the CTDI was the one developed by the US Food and Drug Administration (FDA) and was therefore labeled $CTDI_{FDA}$ and is defined as

$$CTDI_{FDA} = \frac{1}{nSW}\int_{-7}^{+7} D(z)dz$$

where n is the number of distinct planes of data collected during one revolution, SW is the nominal slice width (in mm), $D(z)$ is the dose distribution, and z is the dimension along the patient's axis. For axial (non-spiral/helical) CT scanners and spiral/helical CT scanners with a single row of detectors, $n = 1$. For multislice CT scanners, n is the number of active detector rows (e.g., $n = 64$) during the scan.

This definition, which was accepted by the International Electrotechnical Commission (IEC) [5], is good for all shapes of dose distribution curves $D(z)$ that are emitted by CT scanners. With the $CTDI_{FDA}$, only 14 sections of 7 mm thickness could be measured, and so another dose index, the $CTDI_{100}$, extended the length of the scan measurement to 100 mm. The index is given by the equation

$$CTDI_{100} = (1/nT)\int_{-50}^{50} D(z)dz$$

where nT is the nominal collimated slice thickness.

The next major change in the CT dose descriptor was the introduction of the weighted CTDI ($CTDI_W$) to account for the average dose in the x-y axis of the patient instead of the z-axis, and it is expressed as follows:

$$CTDI_W = (1/3)(CTDI_{100})_{centre} + (2/3)(CTDI_{100})_{periphery}$$

In order to consider the dose in the z-axis, yet another dose descriptor was developed. This is the $CTDI_{vol}$, and it can be calculated using the following relationship for spiral//helical CT imaging:

$$CTDI_{vol} = CTDI_W / Pitch$$

For a pitch of 1, the $CTDI_{vol}$ is equal to the $CTDI_{W}$.

9.1.2 The Dose Length Product

The dose length product (DLP) is yet another dose descriptor used in CT dose studies, and reported in the literature and on CT scanners. While the $CTDI_{vol}$ provides a measurement of the exposure per slice of tissue, the DLP provides a measurement of the total amount of exposure for a series of scans. The DLP can be calculated knowing the length of the irradiated volume (scan length) and the $CTDI_{vol}$ using the following relationship:

$$DLP = CTDI_{volume} \times \text{scan length}$$

It is important to note that while the $CTDI_{vol}$ is not dependent on the scan length, the DLP is directional proportional to the scan length. It is not within the scope of this book to describe the details of how to measure the CTDI.

9.2 Factors Affecting Dose in CT

There are several factors that affect the dose in CT including the exposure technique factors, X-ray beam collimation, pitch, patient centering, number of detectors, and over-ranging (also referred to as z-overscanning); particularly important for the operator are pitch, patient centering, and automatic tube current modulation.

First, note that the relationship between the absorbed dose and pitch is as follows:

$$\text{Dose} \propto \frac{1}{\text{Pitch}}$$

Therefore, if the pitch increases, the dose decreases proportionally. Another important factor under the control of the operator is that of patient centering. The patient must be centered in the gantry isocenter for accurate imaging of the anatomy. Inaccurate patient centering (miscentering) degrades the image quality and increases the dose to the patient, especially with the use of automatic exposure control (AEC) in CT. Finally, automatic exposure control (AEC) is now commonplace on CT scanners. AEC uses a technique referred to as automatic tube current modulation (ATCM) to optimize the dose to the patient while maintaining constant image quality regardless of the size of the patient in the *z*-axis, and the attenuation changes in the *x*-*y* axis.

In CT, ATCM refers to the automatic control of the mA in two directions of the patient (the *x*-*y* axis and the *z*-axis) during data acquisition using specific procedures

that take into consideration not only the patient size but also the attenuation differences of the various tissues. The overall goal of ATCM is to provide consistent image quality, despite the size of the patient and the tissue attenuation differences, and to control the dose to the patient compared with manual mA selection techniques. The interested reader should refer to Seeram [4] for a further description of this technique.

While the automatic control of the tube current (mA) in the *x*-*y* axis (in-plane) is referred to as angular modulation, changing the tube current automatically in the *z*-axis (through-plane) is referred to as *z*-axis modulation or longitudinal modulation. When used together, that is, angular-longitudinal tube current modulation, AEC is the result. The use of angular-longitudinal modulation can reduce the dose by as much as 52 % compared to using only the angular modulation technique [6].

The operator must always pay careful attention to the image quality and dose during a CT examination. Image quality includes spatial resolution, contrast resolution, and noise. While spatial resolution depends on geometric factors (such as focal spot size, slice thickness, and pixel size), contrast resolution and noise depend on both the quality (beam energy) and quantity (number of X-ray photons) of the radiation beam. Several mathematical equations have been derived to express the relationship between dose and image quality. For CT operators, the following mathematical expression is important:

$$\text{Dose} = \text{Intensity} \times \text{Beam Energy}/\text{Noise}^2 \times \text{Pixel Size}^3 \times \text{Slice Thickness}$$

As noted by Seeram [4], this expression implies the following about dose and image quality:

(a) To reduce the noise in an image by a factor of 2 requires an increase in the dose by a factor of 4.
(b) To improve the spatial resolution (pixel size) by a factor of 2 (keeping the noise constant) requires an increase in the dose by a factor of 8.
(c) To decrease the slice thickness by a factor of 2 requires an increase in the dose by a factor of 2 (keeping the noise constant).
(d) To decrease both the slice thickness and the pixel size by a factor of 2 requires an increase in the dose by a factor of 16 ($2^3 \times 2 = 2 \times 2 \times 2 \times 2$).
(e) Increasing mA and kVp increases the dose proportionally. For example, while a twofold increase in mA increases the dose by a factor of 2, additionally, doubling the dose will require an increase by the square of the kVp.

9.3 Radiation Protection Considerations

The effects of radiation can be described as being either stochastic or deterministic. Stochastic effects are usually due to a low dose of radiation which could be received over a long period of time. These effects are random, and the likely outcome is cancer

or genetic effects. There is no threshold value below which it is certain that cancer or genetic effects will not occur but doubling the radiation doubles the risk, that is, the relationship is a linear one. Deterministic effects are usually the result of a higher radiation dose usually over a shorter time period. There is a threshold value above which a deterministic effect will happen, and the severity of this effect increases with the amount of radiation received. An example of this would be induction of cataracts following radiation dose. The minimum single dose necessary to produce a progressive cataract would be 2 Gy. Above this threshold, the biological response will increase. Lower doses of radiation will cause the same effect, but the threshold value will be higher. Cataracts develop approximately 8 years postexposure [7]. The purpose of radiation protection is to limit the radiation dose received by patients (and staff) from medical exposures. Radiation dose received can be increased considerably if X-ray equipment is used inappropriately or radiation protection is inadequate.

9.3.1 Need for Radiation Protection in CT

There has been an increase in the use of CT in recent years which has led to a potential increase in radiation burden to the general population [8]. As well as the increase in use within diagnostic CT departments, there has also been a rise in the number of nuclear medicine departments which now utilize CT.

There are two types of CT scanners that are used in conjunction with SPECT—diagnostic quality with a full range of parameters available and low dose/low resolution where the parameters are less flexible or fixed. Low-dose scanners tend to use a much lower tube current (mA) than diagnostic scanners. As radiation dose is directly proportional to mA when peak tube voltage (kVp), scan time, and slice width remain constant, use of a reduced mA can significantly lower the dose received by the patient [9]. It will, however, have an effect on the quality of the resultant image (see Sect. 10.4).

Diagnostic CT scanners used as an adjunct to SPECT can also be used in a low-dose way by selecting parameters similar to those used by scanners designed to operate at a lower tube current. However, even when used for attenuation correction purposes, it is probable that the slice width and other parameters might vary which in turn could increase the radiation burden of the patient.

9.3.2 Legislation

Each country will have its own arrangements for the regulation of radiation in relation to humans. It is beyond the scope of this chapter to consider all regulatory arrangements, and instead we shall focus into one country (UK). Readers from other countries may find the following information helpful as there are many

commonalities between regulatory arrangements of different countries. However, for specific details of any particular country, the reader is encouraged to review the regulation and guidance documents that apply locally.

The Ionising Radiation (Medical Exposure) Regulations 2000 (IR(ME)R 2000) relates to patient safety with regard to radiation dose. IR(ME)R 2000 specifies personnel who are involved in patient safety during medical exposures as the referrer (the person who requests the medical image to be created), practitioner (the person who justifies that the imaging procedure can go ahead), and operator (the person who physically makes the radiation exposure). Different professional groups may take on these roles, and there can be some overlap between duties. For example, a practitioner can also act as an operator, but an operator cannot necessarily act as a practitioner. It is necessary to have the appropriate skill mix so that justification and optimization of the procedure can be performed. This is a fundamental component of radiation protection.

The Ionising Radiations Regulations 1999 (IRR 99) [10] aims to ensure a structured approach to radiation safety by employers. It defines that all radiation doses should be kept as low as reasonably practicable (ALARP principle). IRR 99 specifies that there should be safe working practices covered by local rules and that there should be specific dose limits for both staff and patients.

9.3.3 General Principles of Radiation Protection

The essential factor in radiation protection is keeping the dose to patient, staff, and members of the public as low as reasonably practicable (ALARP). In general, this can be addressed by justification of referrals and optimization of parameters used and good explanation to the patient to ensure compliance and reduced need for repeat examinations. This corresponds with the basic principles identified by the International Commission on Radiological *Protection* (*ICRP*) [11]. The principles are as follows:

1. Justification
   The principle of justification is that a patient will only be exposed to ionizing radiation if that exposure is beneficial to them and that the benefit they receive outweighs the risk from the radiation dose. It is the first step in radiation protection and is reliant on candid clinical information [11].
2. Optimization
   Optimization can be achieved by selection of the appropriate parameters to ensure that the radiation dose administered is ALARP so as to achieve the required image(s). For SPECT-CT, the CT images are often for attenuation correction (AC) so a diagnostic quality image is not required. This lower image quality commands a lower mA, and so radiation burden to the patient is naturally reduced. This explains why on low-dose CT scanners, which are used purely for AC, the selection of CT acquisition parameters is limited or even set to achieve a very low dose.

Table 9.1 Summary of radiation dose limits

	Employees > 18 years (mSv)	Trainees < 18 years (mSv)	Any other person including members of public and employees < 16 years not in training (mSv)
Limit on effective dose (dose to whole body)	20	6	1
Limit to equivalent dose to lens of eye	150	50	15
Limit to equivalent dose for skin	500	150	50
Limit to hand, forearms, feet, and ankles	500	150	50

Adapted from Schedule 4 of the Ionising Radiation Regulations 1999 [10]

3. Dose Limitation

   As stated previously, limitation of radiation dose is applicable to staff and members of the public as well as the patient. While it is essential that the radiation dose to the patient is limited as much as possible, staff also have a duty to ensure good radiation protection measures are in place for themselves and other patients, visitors, or members of the public escorting patients for X-ray procedures.

   The Ionising Radiation Regulations 1999 have imposed dose regulations which must not be exceeded in order to ensure that the risk of cancer induced by exposure to radiation is not at an unacceptable level [6, 12]. Table 9.1 provides a summary of radiation dose limits.

9.3.4 Designated Areas

Within a department utilizing ionizing radiation, there will be designated areas. These areas are classified as either controlled area or supervised area depending on the amount of radiation dose an employee is likely to receive while working in this area. Monitoring of radiation levels should be done regularly to ensure correct designation of the area is maintained.

In a controlled area, the staff member is likely to receive an effective annual radiation dose of greater than 6 mSv. It is necessary that all controlled areas are clearly demarcated and that warning signs are clearly visible. An example of a controlled area would be a CT scan room. Warning signs should indicate that the area is controlled and that access is restricted. A light indicating that X-rays are being transmitted should be visible on all access points to the room, and all external doors should remain locked during the examination.

A supervised area is one where a member of staff is likely to receive an effective annual radiation dose of greater than 1 mSv. An example of a supervised area would be the control room within a SPECT-CT unit. Again, suitable warning signs should

be visible, and although restriction to a supervised area is less stringent than a controlled area, it is often restricted to ensure patient privacy.

9.3.5 Local Rules

Local rules are a set of written instructions provided for use in designated areas. They should provide a clear reference for radiation safety within the associated area and provide information of contingency plans in the event of an accident. Where local rules are applicable, it is always necessary that at least one radiation protection supervisor is employed to ensure adherence to the rules. IRR 99 specifies that an employee should not knowingly expose themselves or others to ionizing radiation greater than that necessary and that they should make proper use of any personal protective devices and shielding available to them.

9.4 Monitoring and Dose Recording for Staff

It is important to monitor staff radiation dose to ensure that exposure is controlled and within dose limits. If doses appear to be higher than expected, then the information can help prompt investigation into underlying reasons. This could lead to further training or review of working conditions and practices. It should also provide evidence in cases of underlying overexposure or accident. Two of the most common ways to measure radiation dose to staff from ionizing radiation are the use of film badges or thermoluminescent dosimeters (TLDs). These are normally issued on a monthly basis, but this might be reduced to three monthly if doses are found to be low [13].

Radiation monitoring badges are usually worn on the front of the torso at waist height. They should be worn at all times during the working day, and in the event of protective clothing being worn, for example, a lead-rubber apron, the badge should be worn underneath the protective garment. TLD badges are approximately ten times more sensitive than film badges. This renders them more susceptible to changes in background radiation. For this reason, control badges tend to be used to allow subtraction of background doses from personal doses.

9.5 Maintenance of Equipment

It is important that equipment is maintained to ensure that it is performing at a level that is fit for intended clinical purposes. Quality control (QC) tests (Chap. 5) should be carried out on a regular basis to monitor performance of the CT system against an accepted standard to ensure consistency. Any faults which are noted from QC

Table 9.2 Variations between a low-resolution CT scanner and a diagnostic CT scanner

	Low-dose/ low-resolution CT scanner	Diagnostic CT scanner
kV	120–140	120–140
mA	1–2.5	50–400
Rotation time (s)	23	0.32–3.0
Pitch factor	1.9	0.625–1.5
Acquired slice width (mm)	5–10	0.5–2
Reconstructed slice width (mm)	5–10	0.5–10
Reconstruction interval (mm)	3.5–10	0.6–10
Matrix	128 × 128 256 × 256	512 × 512

tests should be reported to the employer immediately as should any damage or faults observed during normal working practice.

9.6 Acquisition Parameter Differences: Diagnostic CT and Low-Dose CT in SPECT

CT is a transmission technique as opposed to the emission technique used in the SPECT part of the study. Therefore, unlike the use of radionuclides, the radiation dose is dependent upon the parameters selected. We have already determined that the way in which low-dose CT equipment varies from diagnostic quality CT equipment (by the parameters that are available for selection). Adding further clarity to this, Table 9.2 demonstrates differences between a low-resolution CT scanner and a diagnostic quality CT scanner.

9.7 Room Design: Shielding

Radiation protection considerations for SPECT-CT need to include radiation from the patient (emission) and radiation from the CT scanner (transmission). The amount of shielding necessary will depend upon the CT scanner capability and the parameters used [14]. For instance, when used in conjunction with a low mA scanner, then 3 mm of lead room shielding is considered sufficient. The estimated workload and existing structural shielding (which could also include floor and ceiling if rooms above and below are occupied) will also be taken into consideration when calculating the required thickness of lead shielding. In most cases, CT scanners used in conjunction with SPECT will have a much lower workload than those used for diagnostic CT purposes.

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Chapter 10
Optimal Utilization of Imaging Parameters in CT Scanning: Uses of CT in Radionuclide Imaging – Different Manufacturer Approaches

Lynn Bateman and Joanne Sil

Both CT and SPECT are well-established diagnostic imaging modalities in their own right. CT with its high spatial resolution is suited to the demonstration of anatomical structures and provides morphological detail such as tissue and disease appearance, size, and tissue composition information. This information is useful for localization of disease and, especially when intravenous contrast medium is used, characterization. SPECT on the other hand is more functional imaging modality, reflecting the metabolic status of disease but lacking the anatomical landmarks intrinsic on the CT image.

As both imaging techniques have different capabilities, they can be used together in a synergistic way. The CT component can be used for registration, to determine the geometric relationship of the data from dual modality techniques. This makes localization of disease and its possible spread into the surrounding tissues more accurate than a single modality alone [1]. Fusion of images from the two modalities enables the characterization of tracer uptake on the planar images and improves diagnostic ability.

There are three different pathways which have been explored by the manufacturers of SPECT/CT systems. One utilizes a low-powered, nondiagnostic CT element with the primary objective of acquiring data to enable attenuation corrections to be made to the SPECT information. Another utilizes a high-powered, conventional CT scanner, which provides diagnostic quality images which can be used for attenuation corrections also. These systems have the added advantage of being able to utilize the CT scanner as a stand-alone unit. This can be particularly beneficial in

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providing backup support or indeed increased CT capacity, in locations with only limited standard diagnostic CT facilities. The third is somewhere in the middle, utilizing a cone beam of X-rays and a large area flat-panel detector. Images are constructed from a number of projections as the tube and detector rotate around the patient. This type of scanner is designed for localization and attenuation correction, not to provide CT quality diagnostic images. This type of system may be beneficial for installations which are limited in available space as their footprint is significantly smaller than most systems utilizing fully diagnostic CT elements.

10.1 CT Attenuation Correction (CTAC)

Activity which has accumulated near to the surface of a tissue will generate a stronger signal than the same activity at a deeper level within the tissue in line with the attenuation value of the tissue the gamma energy is being emitted from. The SPECT images are computed based upon the principle that the magnitude of the detected signal is proportional to the attenuation in the object. Therefore, significant decreases in the gamma energy along its path will increase the attenuation of the subsequent tissues. Left uncorrected, this can lead to spatial distortion within the resultant image especially when gamma radiation is emitted through tissues of significant density. This can be likened to beam-hardening artifact which occurs in CT (see Sect. 10.7.3.1).

An area where attenuation correction is commonly necessary is for myocardial perfusion imaging (MPI). When CT is used for attenuation correction purposes, images are typically acquired when the patient is free breathing so that the CT data matches the SPECT data. This renders the CT images which are produced undiagnostic; indeed, they have been acquired for attenuation correction purposes only and not for diagnosis. Interestingly though, due to the high inherent contrast within the thorax, it is often possible to identify incidental findings on the CT image giving them some diagnostic ability. At this time, the usefulness of these images in the detection of extracardiac pathologies or abnormalities is uncertain and further research is necessary.

It should be noted that the CT acquisition does not occur concurrently with the SPECT acquisition, and indeed, the patient couch may have to move between the two scans as, dependent on the system employed, the axis of rotations may be different for CT and SPECT. Also, the use of intravenous contrast agents for the CT study may have an adverse effect on the calculated level of attenuation correction required. The contrast agent within the tissues and vessels will increase attenuation of the gamma rays leading to artifact and complication of attenuation correction process.

An important stage in the attenuation correction process is for the attenuation map produced by the CT element to be corrected to the energy of the radionuclide in use. Typical average beam energies in CT are of the order of 70 keV but, for example, the frequently used Technetium 99m emits 140 keV gamma radiation. This energy difference leads to significant differences in tissue attenuation.

Despite this necessary step, attenuation correction utilizing CT data has many benefits. Compared to methods utilizing radionuclide sources, the CT element

produces the attenuation information very quickly and easily. The data has low noise levels, and it also provides anatomical information.

The CT scan will obviously carry a dose burden for the patient; the magnitude of this will vary dependent on the anatomical area covered. As with every exposure to ionizing radiation, the dose should adhere to the ALARA (as low as reasonably achievable) principle, and the acquisition parameters should be matched to the clinically justified need; for example, if the image is primarily required for attenuation correction purposes, it should be possible to acquire it with relatively low dose.

10.2 The CT Element

CT scanners are designed to be menu driven. The operator will have the ability to select from a collection of preprogrammed protocols based on anatomy and patient size. Ideally, the protocols will have been established by collaboration between the applications specialist, radiology staff (radiographers and radiologists), and a radiation physicist (medical physics expert) such that the quality of the images is sufficient to provide the diagnostic information with which to answer the clinical question, at the minimum possible radiation dose.

As each patient is unique, the parameters of such CT scanning protocols may be modified by the operator to optimize image quality and dose to the patient. The choice of parameters selected involves various trade-offs between image quality and radiation dose, and these need to be understood before the operator can make a decision on how to best use the parameters for optimization. Excessive radiation dose in CT imaging may not be obvious by observation of the images. As CT is a digital modality, the image will not appear to the observer to be darker with increasing radiation dose, the imaging system utilizes pre-set viewing levels depending upon the protocol, and increasing the scanning parameters will decrease the level of noise within the image. There is a point when there will be no visible changes to the image as the dose increases, so it is relatively easy to use excessive exposure factors for examinations, with no clinical benefit.

We will discuss the properties of the individual parameters available and how changing their value affects image quality and patient dose. Information is broadly separated into properties of the X-ray tube, properties of the scanning protocol and/or the system, and finally properties of the image processing. These categories are somewhat arbitrary, and exact details can be scanner manufacturer dependent. It should be stressed that the reader should acquaint themselves with the methods and terminology used by their scanner manufacturer.

We begin, however, with a brief overview of a few important definitions.

10.2.1 What Does a CT Image Represent?

A CT image is a visual interpretation of the variations in linear attenuation coefficient (μ) [MU] of the objects in the area scanned. As the X-ray beam travels on its path

through the body, it is attenuated by differing amounts dependent on the tissues encountered. The CT system rotates the X-ray beam and detectors around the patient acquiring data from a large number of projections as it goes. It is able to combine data, from these projections, and compute an image based on the distribution of differing attenuations (see Sect. 10.7).

In order to visualize the image and to allow differentiation of tissues which have very similar attenuation characteristics, the value of μ which has been determined for each picture element, pixel, is converted to a number on the Hounsfield scale, after the creator of CT, Godfrey Hounsfield. This scale defines the value of two materials and relates other materials to them. The Hounsfield unit (HU) of water is defined as 0 HU, and air is defined as −1000 HU. All other materials are determined from the formula

$$CT\# = \left(\frac{\mu_{\text{Tissue}} - \mu_{\text{Water}}}{\mu_{\text{Water}}} \right) \times 1{,}000$$

Typical values of Hounsfield numbers of various tissue types are shown in Table 10.1. These values are frequently referred to as CT numbers, CT#.

It should be noted that μ is dependent on the energy of the X-ray beam, and therefore, the CT# of a material may vary with beam energy.

The accuracy of these values may depend on the type of scanner in question. If the CT element is primarily intended for CTAC use, there is less importance placed on the accuracy of indicated CT number, and they may vary from the expected value by ±40 HU.

10.2.2 What Is Image Noise?

The noise in a CT image is defined by the standard deviation (SD) of the CT numbers within the image. There are various factors which contribute to the overall level of noise in the CT image including the number of photons contributing to the image,

Table 10.1 Typical values of CT number

Tissue/material	Typical CT#
Air	−1,000
Adipose	−100
Water	0
Muscle	40
Kidney	20–40
Brain	55
Liver	50–70
Polymethylmethacrylate (PMMA, “Lucite,” or “Perspex”)	140
Cortical bone	600 to 2,000+

inherent system noise, efficiency of the detectors, X-ray tube performance, and reconstruction algorithms. The most significant source of noise arises from the number of photons used for the image, which is primarily controlled by the current applied to the X-ray tube (see Sect. 10.5.2). Decreasing the number of photons used increases the noise significantly as the noise is related to the inverse square root of the tube current. As the patient dose is also directly affected by the tube current (mA), there has to be a balance between image noise and acceptable patient dose.

$$\text{Noise} \propto \frac{1}{\sqrt{\text{mA}}}$$

If an image is very noisy, differentiation of regions of low contrast difference can be difficult or impossible, rendering the image useless and the entire radiation dose unnecessary.

Quantification of the level of quality in the image can be obtained using the quotient of image signal (CT#) and inherent noise (SD) in any particular region. The signal-to-noise ratio, SNR, increases with increasing tube current as the fluctuations in the signal level decrease.

It is not possible, however, to produce images with no inherent noise as the production of X-rays is a random process. One area of the beam will contain more photons than another; hence, the distribution of photons over the detector is also random. This causes fluctuations in the signal level in any particular pixel.

Other contributions to the intrinsic noise in the system arise from the multitude of mechanical and electrical components used in CT scanners. However, the magnitude of such noise is low compared to other sources.

A significant contributor to the non-intrinsic elements of noise arises from the choice of image processing algorithm, with those designed for visualizing boney details adding very significant noise levels to the image. These are discussed further in Sect. 10.7.

10.3 How Is Radiation Dose Described?

It is important for the reader to have an appreciation of the differences between the dose metrics which are used to describe radiation dose. This section outlines the most important ones used in CT imaging.

10.3.1 Absorbed Dose (D)

This is the amount of energy from radiation exposure that is absorbed into the tissue per unit mass. It is measured in units of joule/kilogram and has the special name of the gray (Gy). The absorbed dose is the average dose across a tissue or organ and is the dose used for dosimetry purposes.

10.3.2 Equivalent Dose (H)

The effect on the tissue absorbing the radiation can vary greatly for the same absorbed dose depending on the nature of the radiation absorbed. Equivalent dose takes into account the effect of the specific radiation type on an organ or tissue and is a measure of the risk of exposure to ionizing radiation. Radiation types are each assigned a weighting factor, and for X-rays, this is equal to 1.

$$H = D \times \mathrm{W_R}$$

where H is the equivalent dose, D is the absorbed dose, and $\mathrm{W_R}$ is the radiation weighting factor.

Equivalent dose is measured in sieverts (Sv).

10.3.3 Effective Dose (E)

Sensitivity to ionizing radiation is different for different tissue types with some tissues being more susceptible to the effects of ionizing radiation than others. Tissue weighting factors (W_T) are used for to account for the sensitivity of tissue types in a similar way to the weighting factors used for different radiation types when determining equivalent dose. The tissues (or organs) which have been assigned tissue weighting factors are listed in Table 10.2. The tissue weighting factors are multiplied by the equivalent dose each tissue (or organ) receives during the scan. These products are then added together to give a total which is the effective dose, E, delivered by the scan.

Effective dose is the metric that is best used to describe the dose delivered to a patient as it represents the dose and its effect to the whole patient, not just to an organ. Effective dose is also used to compare the doses delivered by different imaging modalities.

Table 10.2 Tissue weighting factors [2] $E = H \times W_T$

Tissue	Tissue weighting factor W_T	Sum of W_T
Bone marrow (red), colon, lung, stomach, breast, remainder organs[a]	0.12	0.72
Gonads	0.08	0.08
Bladder, esophagus, liver, thyroid	0.04	0.16
Bone, brain, salivary glands	0.01	0.04
Total		1.0

[a]Adrenals, extrathoracic (ET) region, gall bladder, heart, kidneys, lymphatic nodes, muscle, oral mucosa, pancreas, prostate, small intestine, spleen, thymus, uterus/cervix

10.3.4 Computed Tomography Dose Index (CTDI)

In a single X-ray projection, the radiation intensity at any point decreases as the attenuation in its path increases.

As, in CT scanning, the X-ray tube is rotated around the gantry, this therefore provides a relatively uniform region of radiation dose across the air of the gantry aperture, when considered over a complete tube rotation. Even though the X-ray fan beam is well collimated, there is still some broadening in the *z*-axis; the dose profile does not resemble a square wave. This makes absolute determination of the dose level difficult. In order to quantify the level of absorbed dose (to air) over a single tube rotation, the concept of computed tomography dose index, $CTDI_{100}$, was introduced. $CTDI_{100}$ (where the subscript indicates the detection length) represents the total dose delivered per rotation and is measured free in air at the center of the gantry, using a 100 mm long detector such that, at least for traditional X-ray beam widths, the complete dose profile falls over the detector length. $CTDI_{100}$ has units of milli Gray, mGy, and is a measure of the absorbed dose to air. As soon as any additional attenuation is introduced, a modified dose metric, weighted CTDI, $CTDI_{W,}$ is required. This is again defined over 100 mm but this time in either a 16 cm (head) or a 32 cm (body) polymethyl methacrylate (PMMA, Perspex, Lucite,) phantom in an attempt to recreate similar attenuation patterns to those in a patient. The calculation uses 1/3 of the CTDI at the center and 2/3 of the CTDI at the periphery of the relevant phantom. If the scanned area is small, for example, as in head scanning, the dose distribution is approximately uniform throughout the scanned region. As the anatomy (attenuation) increases, for example, in the abdomen, there is a dose gradient introduced, and the dose at the patient's skin is approximately twice the level of the dose in the center of the patient. Hence, $CTDI_W$ attempts to provide the average dose to polymethyl methacrylate per rotation.

This value is useful for contiguous (abutting) scans; if this is not the case, a further definition modification is required so the average dose over the scanned volume is described: CTDI volume, $CTDI_V$. This is simply the $CTDI_w$ divided by the pitch (see Sect. 10.6.5). It is likely that this value will be displayed by the scanner. The value of $CTDI_V$ gives an indication of the average dose delivered for a scan sequence but is not a good indication of individual patient doses.

10.3.5 Dose Length Product (DLP)

In order to gain an improved appreciation for the dose that a particular set of scan parameters will deliver, and to allow comparisons of doses for a specific examination on different CT scanners, the length of the scan also needs to be accounted for. The dose length product, DLP, is defined as the $CTDI_V$ multiplied by the scan range in centimeters and has units of mGy · cm. Along with $CTDI_V$, DLP is likely to be displayed on the scanner console. It should be noted that it is possible that scans can be

performed which deliver identical values of DLP, but dependent on the anatomy covered, they may deliver different effective doses to the patient due to the coverage of tissues which have differing sensitivity to radiation as disused earlier.

10.4 CT Scanner Development

Since its inception in the 1970s, CT scanning and scanner technology have undergone large developmental changes. The early days of acquisitions utilized combinations of narrow "pencil" X-ray beams and single detectors which acquired data by making measurement in one location then moving both tube and detector step wise in a grid-like fashion over the area of interest (analogous to rectilinear scanners in radionuclide imaging). Technology then progressed through various angles of fan-shaped X-ray beam and a bank of detectors located opposite to the X-ray tube, rotating together around the patient. The most recent developments in scanner technology have revolved around the number of detector elements which are utilized in the *z*-axis (the axis perpendicular to the gantry). These so-called multi-detector CT scanners have dominated the market for some time now.

10.4.1 Scan Modes

Historically, CT took the form of incremental axial or *step-and-shoot* scanning. This involved the patient remaining in a stationary position, while the X-ray tube produced X-rays and was rotated 360° around the gantry and hence the patient. The X-rays' incident on the opposing detector(s) during the rotation was used to form the data which was processed to produce one axial slice. Between obtaining each slice, there was an inter-scan delay while the tube returned to the home position and the table and patient incremented into the gantry in readiness for the next slice to be acquired. Technology improvements lead initially to a reduction in this delay and then on to the ability to have the X-ray tube and detectors continuously rotating around the patient, without the need to return to a home location. From here, it was a short step to allowing the patient to be moved through the gantry as the X-ray tube and detectors rotated around them, giving rise to helical scanning. This technological increase allowed typical scan times for the abdomen and pelvis to decrease significantly from the 20 minutes it frequently took to acquire axial slices of around 10 mm to imaging the area in a matter of seconds.

Helical scanning not only has the obvious advantage of offering a vast reduction in scan time, but also, as it collects a volume of data, allows this data to be viewed in different planes, for example, coronally.

10.4.2 Image Increments

When scanning in incremental mode, it is usual to acquire direct axial slices which are contiguous or abutting. Contiguous axial slices ensure that there are no gaps between images and help to ascertain that there is no loss of anatomical detail in the resultant images. As we discussed earlier (Sect. 10.4.1), however, incremental scanning of larger areas of anatomy takes longer than scanning the same area helically. For scanning of the trunk, this extended scan time can cause problems due to breathing artifact. For diagnostic scanning, patients are asked to hold their breath for each axial acquisition to reduce movement artifact caused by the breathing cycle. Understandably, not each breath the patient takes will be of the same volume, and this is often a cause of misregistration between patient anatomy and image anatomy even with contiguous slices.

For diffuse disease, it is not always necessary to acquire contiguous axial slices. An example of this would be when employing high-resolution CT to image for bronchiectasis or other diffuse lung disease.

As described, helical scanning acquires a volume of data rather than discrete axial slices. While it is still commonplace to view the reconstructed images in a contiguous axial manner, it is now possible to manipulate the image data. Hence, image data can be utilized to produce images representing different planes (multi-planar reformatting or MPR), to provide 3D representations and also to reconstruct the data into slices of different widths and at different intervals. For studies requiring high-quality reformatted images, such as vascular studies, it is possible to reconstruct images which overlap or to reconstruct the data into multiple formats utilizing different algorithms, without increasing the dose to the patient. The limiting factor for this may be the increase in post-processing time necessary for the reconstruction; however, it is possible to include these different reconstructions into the scanning protocol which is programmed into the scanner, so they are carried out automatically and will be ready for the images to be reported upon.

10.4.3 Multi-detector CT

The latest significant development in CT scanners has come from the expansion of the X-ray beam in the *z*-axis and corresponding reconstruction algorithms to deal with an X-ray beam which is expanding from a fan shape tends toward a cone (see Sect. 10.7.2). This means that the detectors are comprised of a bank of elements in the *z*-axis which can be used to produce individual slice data, or their signals combined to produce images representing thicker sections of the patient, but which still retain the high spatial resolution provided by collecting data over a small area. Dependent on the scanner, it is currently possible to define a scan which uses an X-ray beam width of 2 mm (at the isocenter) which

is detected by a bank of 4 × 0.5 mm detector elements (slices) to one utilizing a beam width of 160 mm detected as 320 × 0.5 mm slices.

This development further decreased the time required to produce a scan, opening up more and more clinical possibilities. It is now possible to image patients quickly enough so large volumes of the patient are scanned with very high resolution in a single breath hold. The very latest scanners are capable of imaging an entire organ (e.g., heart) in a single rotation of the X-ray tube and detector, which takes a fraction of a second (of the order of 0.23 s).

10.5 Properties of the X-Ray Beam

This section defines tube voltage and current, rotation time, and acquisition slice width and describes the effects they have on the patient dose and resultant image quality.

10.5.1 Tube Voltage (kV)

When a potential difference is applied across an X-ray tube, a spectrum of X-rays are produced (as described in Chap. 3). The maximum energy possessed by the X-ray beam is defined by the magnitude of the potential difference, and this defines the peak kilovoltage (kVp). The term "kV" is often used to describe the tube voltage.

CT scanners typically have a range of 3 or 4 tube voltage settings between 80 and 140 kV which can be used clinically. The choice of kV may be restricted by the system dependent on the protocol selection. However, it is currently the case that the majority of scans are performed at 120 kV.

The primary effects of adjusting the kV are to alter the contrast of tissues within the image (as the linear attenuation coefficient varies with energy) and to adjust the number of X-rays produced (if tube current and rotation time are constant). Reduction of the tube voltage will result in increases in the tissue contrast but also in increased image noise due to a reduction in the number of X-ray photons. Conversely, as the kV, and therefore the average energy of the X-ray beam is increased, it becomes more penetrating; however, the overall image quality may suffer due to the decrease in image contrast. It is important that a correct balance of kV and mAs (see Sect. 10.5.2) is obtained for each patient. The kV needs to be set such that the resulting beam has sufficient penetrating power for the attenuation levels for the anatomy in question, then the mAs is adjusted to ensure sufficient numbers of X-ray photons are produced. It should be possible to use a lower kV for pediatric scans due to their reduced size and benefit from the increase in penetrating power for imaging of larger patients. In this way, the kV can be used as a tool for image optimization.

It is a common practice in CT imaging to use contrast media to enhance visualization of areas with low inherent contrast differences. It is possible to use the administered contrast media in combination with the kV to optimize patient doses and image quality.

It should also be noted that changing the kV may influence the displayed CT number for any particular material due to variations in the response of the linear attenuation coefficient, μ (mu), with varying energy.

10.5.2 Tube Current (mA)

To produce an image with an acceptable level of quality, the dataset must contain sufficient numbers of data points. This means that sufficient numbers of X-ray photons must reach the detectors. While the applied voltage supplies the X-ray beam with its penetrating power, the applied tube current influences the intensity of X-rays produced and as such is directly proportional to the radiation dose received by the patient.

Dependent on the scanner manufacturer, it may be either the tube current (mA) or the product of the tube current and the exposure time, the mAs, which is defined in the scanning protocols.

As described before (see Sect. 10.2.2), the tube current has a direct effect on the noise level in the image, that is, increasing in the tube current reduces the noise and hence improves the image quality. However, it is important to remember that this also results in a higher radiation burden to the patient.

In order to optimize the radiation used, it is important that the produced images have a noise level which is appropriate for the imaging task in hand. As less noise can be tolerated when imaging to visualize small, low contrast details, a proportionally higher tube current is required than if you are imaging to visualize gross changes.

Due to the inefficiencies in the production of X-rays, large amounts of heat are generated by the X-ray tube. The heat produced is proportional to the total number of photons produced, that is, to the mAs. Modern CT tubes have very good heat capacities and cope well, producing very high photon fluxes with high currents and very short rotation times, and are capable of sustaining such factors over many, many rotations. It is as a result of the increases in heat capacity of the X-ray tubes that other elements of CT technology have been able to move forward. Scanners are now capable of acquiring images of large volumes of a patient very quickly or of scanning at speeds to effectively avoid issues arising from patient motion. Such improvements have increased the flexibility of CT scanners and have extended the number of conditions for which CT scanning is indicated.

10.5.3 Rotation Time (s)

This is the time required for the X-ray tube and detector bank to rotate 360° around the patient. Most modern CT scanners can perform a single rotation in less than 0.5 s. Such rotation times, along with increases in the X-ray beam width in the z-axis (along the length of the patient) lead ever closer to imaging which has very high

temporal as well as spatial resolution. Increased rotation times not only allow faster scanning, but a greater volume of tissue can be imaged in a single breath hold, hence decreasing problems caused by differential organ location between acquisitions. It is also possible for some patients, for example, pediatrics, to be scanned quickly enough to remove the need for sedation.

In radionuclide imaging, it may not always be beneficial to perform sub-second CT imaging. Due to the time required to acquire the radionuclide scan dataset, the patient has to breathe freely and hence organs move during the acquisition. If the CT data is to be utilized for attenuation correction purposes, this too should be acquired with the patient free breathing. Therefore, SPECT/CT scanners with high-power CT elements can perform the CT scans using a slower rotation time of typically 1.5 s, and systems with low-power CT elements may take 20–30 s.

10.5.4 Acquisition Slice Width

The slice width that is selected for scanning purposes is referred to as the acquisition slice width. This slice width is a true or primary parameter in that it is selected prior to scanning and cannot be changed post scan. How the images are viewed is determined by the reconstructed slice width which may be the same or greater than the acquisition slice width and as a secondary parameter may be changed post scan. Further discussion on the reconstructed slice width is given in Sect. 10.5.6.

In single slice scanning, typically, the acquisition and reconstruction slice widths are the same and represent a balance between obtaining adequate image noise levels in a reasonable time period while being mindful of the tube loading capacity. Due to the improved tube loading capacity already mentioned and the increased coverage possible with multi-slice scanning, it is a common practice to image almost universally with narrow slices.

Spatial resolution, contrast resolution, partial volume artifacts, and radiation dose are all influenced by the choice of acquisition slice width. Spatial resolution is the ability to detect two small, discrete, but closely situated objects. It stands to reason then that better resolution of small objects will be achieved with a narrower acquisition slice width, due to the fact that each pixel in the image represents the average attenuation of the area it represents. However, by selecting a narrow acquisition slice width, the number of X-ray photons detected by each detector element will be reduced. As mentioned before, the number of photons contributing to the image affects the signal-to-noise ratio, and so with a reduced number of photons, the noise level of the image will be increased, and subsequently, contrast resolution will decrease. This demonstrates that there is an inverse relationship between spatial resolution and contrast resolution. In order to maintain a suitable level of noise in each slice, the tube current, and hence the dose, is likely to be greater than that required for imaging the anatomy with thicker slices.

10.5.5 Geometric Efficiency

It is important that the unattenuated X-ray beam profile is consistent over all of the detector elements in the *z*-axis, so that the system can determine actual attenuation changes in the anatomy being imaged. Hence, the nonuniform edges (penumbra) of the X-ray beam have to be discounted during image production, and the radiation in this region contributes only to patient dose. This issue becomes more significant with multi-section detectors than it was with single detectors. It typically means that the total beam width in the *z*-axis is 2–3 mm wider than the active area of the detector. However, as the beam width in the *z*-axis increases, the dose associated with the beam penumbra becomes less of a significant proportion of overall dose, and the geometric efficiency is said to increase.

Geometric efficiency also improves with the number of slices simultaneously acquired. A small number of thin simultaneous slices requires a small nominal collimation and may be associated with a geometric efficiency of the order of 30 %. As more simultaneous slices are acquired, the nominal collimation increases, and the geometric efficiency tends toward 100 %. This is because the constant size of the penumbra becomes a smaller proportion of the overall beam width. Hence, it is prudent to utilize the maximum collimation consistent with the required images to optimize patient doses.

The phenomenon is known as "over-beaming," and manufacturers are beginning to introduce additional collimation elements which prevent this penumbra dose reaching the patient.

10.5.6 Reconstruction Slice Width

It is possible to combine data from consecutive detector elements, or rotations, in order to produce reconstructed slices of widths that are greater than those used for the acquisition. This may help to balance resolution and dose, but dependent on the scanner design, acquisition, and image reconstruction parameters used, there may be limiting factors imposed. These are discussed in Sect. 10.7. Scan protocols can be defined so they automatically process a dataset in a number of different ways, for example, narrow slices and sagittal views for evaluation as well as thicker axial slices for storage.

A potentially significant issue associated with imaging with very narrow slices is the volume of data that is produced. A single thorax, abdomen, and pelvis scan using 0.5 mm slices could return 1,500 slices. This corresponds to approximately 750 MB of data storage, assuming only one set of images are reconstructed. Clearly, data storage will very quickly become an issue so careful consideration should be given to the long-term storage requirements.

Increasingly, clinicians are moving away from viewing CT image datasets as a series of sequential slices, to viewing the data in 3D, or as multi-planar reconstructions (MPRs) in coronal and/or sagittal orientations (Sect. 10.7.2 reviews image reconstruction techniques).

10.5.7 Beam Filters

Typically, two types of filter are used in CT scanners. Firstly, a flat copper or aluminum filter is used to increase the average energy of the beam by removing the low energy elements of the spectrum. Secondly, a "bow-tie" filter is used to shape the X-ray beam so that it is of uniform intensity at the detectors. These bow-tie filters are designed to modify the X-ray beam such that once it has passed through a spherical uniform attenuator, the intensity at the detector face is approximately constant. Clearly, this is a process of compromise, as a real patient will modify the X-ray beam differently to a uniform spherical object and indeed differently depending on the projection and thickness (attenuation) of the patient. Modern scanners attempt to overcome some of the problems this phenomenon causes by utilizing different filters dependent on the field of view or the anatomical protocol selected; hence, care should be taken that an appropriate protocol is selected when setting up each scan.

10.6 Properties of the Protocol and/or Scanner

In this section, we are going to look at the properties of a scanning protocol. In reality, parameters will be built into a ready-made scanning protocol which has been put together so that it is fit for purpose for a particular category of scan. However, it is best practice to match the scanning parameters, and hence the radiation dose, to both the patient and to the clinical indication in hand. It should be possible to use lower exposure parameters for smaller patients (e.g., pediatrics) or when looking for gross anatomy.

Understanding of how the selectable parameters influence the dose and the image quality will allow the operator to make informed decisions about the required combinations.

10.6.1 Scan Projection Radiographs (SPR)

These are localizer scans used primarily in order that the operator can define the required scan position and range on the patient. They are projection images acquired with the X-ray tube in a stationary position and the couch moving the patient through the gantry. Dependent on the scanner and the procedure to be carried out, they are

obtained for either anterior-posterior (or posterior-anterior) or lateral projections. They are generally low-dose procedures (particularly when compared to the dose for the CT scan) and may actually contribute to patient dose reduction as they form the basis of the tube current settings utilized by the automatic tube current modulation systems described in Sect. 10.6.7.

There are manufacturer-specific names for these scans including the terms "scout view," "topogram," and "scanogram." It is typical for the SPR(s) required to be included as part of the scanning protocol installed on the CT scanner. Although the length of the SPR tends to be preset within a scan protocol, it is possible to increase or decrease this length as necessary. The system should display the image in real time and should provide a mechanism to stop the SPR once the desired range has been covered to ensure that patient radiation dose is kept to a minimum.

10.6.2 Field of View

When discussing field of view (FOV) in relation to CT, it is necessary to appreciate the difference between the two different types that are used, namely, acquired FOV and reconstructed FOV (see Fig. 10.1). Acquired FOV is sometimes referred to as scan FOV or calibrated FOV. Reconstructed FOV is alternatively referred to as display FOV. In this chapter, the terms acquired and reconstructed will be used for consistency.

Acquired FOV is the actual area within the gantry from which data is attained and is determined by the angle of the fan beam and therefore represents a primary parameter.

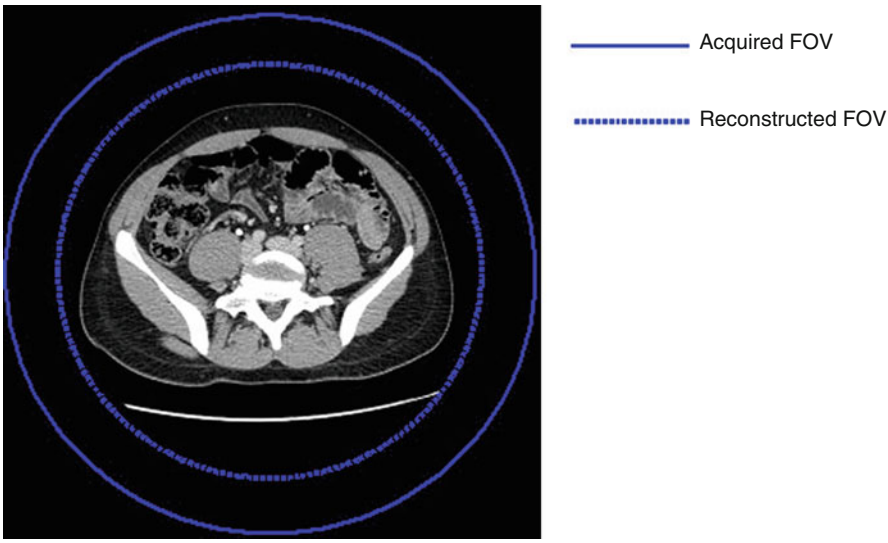

Fig. 10.1 Difference between the acquired and reconstructed field of view

The wider the angle of the beam, the greater the number of detectors that will be irradiated and the larger the acquired FOV will be. Most manufacturers arm their CT scanners with a choice of predetermined acquired FOVs from 200 mm for a pediatric or head scan to as large as 500 mm for body scanning. It is important to use the correct field of view to ensure that the required anatomy is demonstrated while keeping the patient radiation dose to a minimum. As described in Sect. 10.5.7, the choice of bow-tie filter will depend on the FOV selected. Acquired FOV is directly proportional to radiation dose so the larger the FOV selected the larger the dose to the patient.

Clearly, as the size of the acquired FOV determines the extent of the radiation beam, this will define the maximum size of the reconstructed FOV.

The size of the reconstructed image is the reconstructed FOV (see Fig. 10.1) which is a secondary imaging parameter and can be determined at the time of scanning or retrospectively from the raw data.

Although changing the reconstructed FOV does not have any consequence with regard to radiation dose, it is important to select an appropriate size FOV to optimize image quality. The number of pixels used to display the image is set by the image matrix. The best image quality will be realized when the maximum number of pixels are utilized within the image. Looking at Fig. 10.2, it can be seen that if the reconstructed FOV remained the same as the acquired FOV, there would be a large number of pixels being used to display air around the patient and so reducing the number of pixels displaying the useful information from the patient. If the reconstructed FOV is reduced to display more of the patient anatomy, then more pixels are utilized in the image effectively making the pixels smaller (see Fig. 10.2). In this way, spatial resolution in the image is improved.

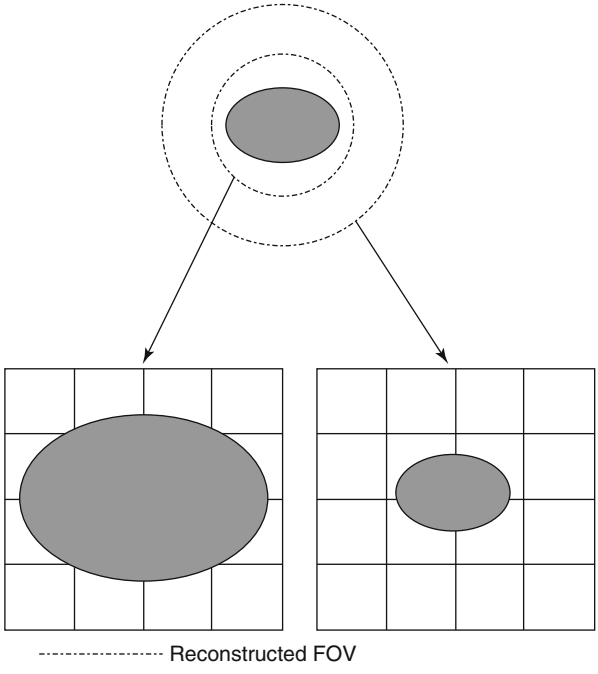

Fig. 10.2 Demonstration of how a smaller reconstructed FOV utilizes more of the pixels to display useful information from the patient in the displayed image compared with a larger reconstructed FOV which uses pixels to demonstrate information outside the patient

It is important to note that utilizing a large reconstructed FOV and hence a smaller number of pixels to represent the patient data will cause the image to appear smaller on the monitor screen. Using the systems' magnification tools would only act to magnify the pixels used to display the image and will not result in improved resolution.

10.6.3 Patient Positioning

It is usual for the patient to be positioned on the couch in such a way so the region of anatomy being imaged lies at the center of the CT scanner gantry, that is, at the isocenter. This may not always be possible due to patient habitus or limitations that the SPECT/CT system might have on available scan range. If patients are not located centrally with respect to the isocenter, there is a potential for increased doses particularly in the regions which are closer to the X-ray beam.

10.6.4 Scan Length

As previously mentioned, before a CT scan is carried out, it is usual to obtain a SPR view in order to define the scan location and extent (see Sect. 10.6.1).

The scan length is determined by the area of anatomy being scanned. There is obvious importance to include all the relevant anatomy within the SPR and consequent scan range. However, it is also important to remember that any unnecessary increase in the scan range will also have dose implications for the patient; increase in scan length increases the total dose that the patient will receive. Therefore, the scan length should only be as long as necessary to include the anatomical area required.

Again, it is probable that an initial indication of the scan length will be included as part of the scan protocol and it will be necessary to adjust the required scan range for the patient in question. During the CT scan, indicative images are displayed in real time on the monitor (prior to the reconstructed images being available). Although these are of lower quality than the reconstructed images, they do allow the operator to monitor the progress of the scan enabling them to terminate it if the desired anatomical range has been covered. Conversely, it is also possible to increase the scan range at the conclusion of the scan if it is considered that a larger anatomical area needs to be covered.

10.6.5 Helical Pitch

Helical CT scanning involves continuous movement of the patient couch through the gantry during the scan acquisition. The X-ray beam path describes a helix across the patient, hence the name helical (also referred to as spiral scanning). This allows either a larger volume of the patient to be imaged within the same time as that

required for axial scanning or imaging the same volume but more quickly. Together, both of these options allow much more flexibility to the operator and increase CT's suitability for different clinical indications.

Helical pitch is the term used to describe the relationship between the distance moved by the patient couch per rotation of the X-ray tube and the X-ray beam width in millimeters.

$$\text{Pitch} = \frac{\text{Couch Travel per tube rotation}}{\text{Total beam width}}$$

It is important that the denominator of this equation represents the entire X-ray beam width, so in multi-slice scanning, this equates to the nominal acquisition slice width multiplied by the number of simultaneous slices, for example, $0.6 \times 64 = 38.4$ mm or $0.5 \times 16 = 8$ mm. When the pitch is equal to 1, for each rotation of the tube, the patient has incremented the distance of the X-ray beam. This best equates to contiguous acquisition frequently used in incremental scanning.

Selecting a pitch of less than 1 indicates that the patient moves a distance which is less than the X-ray beam width for every tube rotation. This gives rise to oversampling which increases the resolution in the z-axis, but it may also give rise to an increase in radiation dose. Using a pitch of greater than 1, and so increasing the distance between samples, may decrease the radiation dose to the patient but decreases the resolution in the z-axis. The selection of pitch, as with other parameters is a trade-off between image quality and radiation burden to the patient.

If the tube current, rotation time, and applied voltage are kept constant on single slice systems, as the pitch increases, the dose to the patient decreases as the coils of the helix become further apart. In turn, this reduces the spatial resolution and increases partial volume effects in the resultant images. The noise level in the images is not affected by changes in pitch. Increasing pitch will also have the effect of increasing the effective slice with (due to the methods for image reconstruction discussed in Sect. 10.7.2). The situation when considering multi-slice scanners is not always straightforward, again due to the image reconstruction methods (see Sect. 10.7.2). Some changes in the scanning parameters (e.g., pitch) made by the operator may lead to changes in image quality unless other parameters are also adjusted. In an attempt to minimize such effects, the concept of "effective mAs" has been introduced, rather than simply mAs. The operator defines the effective mAs in the scan protocol, and the system automatically adjusts the mA aiming to maintain noise levels as the pitch varies. Using this parameter means that the dose and noise levels are independent of the pitch used.

10.6.6 Gantry Tilt

Tilting of the gantry may be possible on diagnostic CT scanners, but this is not always the case with hybrid scanners. By definition, gantry tilt is the angle that the

X-ray tube has been moved from the vertical position. The range of gantry tilt available for most modern scanners is between + and −25°.

Angulation of the gantry is based on clinical decision. Traditionally, in the days of incremental scanning, the gantry was often angled to gain a specific plane through an anatomical area. An example of this might be angulation to Reid's baseline for scans of the brain. This avoided scanning directly through the dense petrous bone which was frequently the cause of beam-hardening artifact (see Sect. 10.7.3.1) and degradation of the resultant images. Of course, this involved scanning through the orbits of the patient and more specifically the highly radiosensitive lenses.

Current practice now is to avoid scanning through the lens to reduce the induction of cataracts from the radiation dose received. The change in baseline to that of a low radiation baseline where the baseline is angled to the supraorbital margin either by patient positioning or angulation of the gantry has been made possible by the progress in scanning techniques of modern scanners. So for contemporary scanners, the protocols tend to incorporate gantry tilt to avoid or reduce radiation dose to anatomical areas with a high radiosensitivity.

10.6.7 Automatic Tube Current Modulation (ATCM)

Historically, a scan would be acquired using the tube current setting that was most appropriate for obtaining adequate image quality of the thickest or most attenuating region of anatomy within the scan range. It was likely to be defined in the protocol and the level based on a "typical" patient. However, as not all patients are "typical" and different areas of the body consist of tissues of differing densities, or have different inherent contrast, it is now possible for CT systems to automatically adjust the tube current to accommodate these differences. For example, the thorax contains low-density lung tissue containing air which causes less attenuation to X-rays than the higher density tissue of the liver found in the upper abdomen. If the same mA was selected for both areas, more X-rays would reach the detectors and contribute to the image from the thorax than the abdomen. As the mA would have been selected for the more attenuating abdomen region, the excess of X-rays contributing to the images of the thorax purely serve to increase the radiation dose to the patient. Hence, a lower mA could be utilized through the thorax while still maintaining sufficient levels of image quality. Additionally, the large range of densities within the thorax also means that it has good inherent contrast compared to the abdomen where the tissue densities are all very similar. This implies that adequate imaging of the thorax may be achieved with proportionally fewer X-rays than required for the abdomen.

We have already discussed (see Sect. 10.2.2) how increasing the mA affects the number of X-rays reaching the detectors, increasing the signal and consequently the contrast resolution in the resultant image. What we are now considering is the reverse, in that it would be possible to reduce the mA in areas of the body that are less dense and cause less attenuation of the X-ray beam yet still maintain the signal-to-noise ratio and contrast resolution.

Automatic tube current modulation systems (ATCM), also termed "automated mA" or "mA modulation," are based on the principle that image noise is determined by the number of X-rays reaching the detectors. The aim of these automatic systems is to adjust the tube current so that image quality is maintained at a constant level, which is predefined in the protocol or by the operator, while keeping the dose to the patient as low as possible. The operator influences the settings by either defining an acceptable level of noise, or by setting an effective mAs level, that is, the mAs which they would select for that scan without the use of automatic modulation. There are three main methods employed to achieve this which are *x*-*y*-axis (angular) modulation, *z*-axis modulation, and real-time "on the fly" modulation in all three axis. The names given to each modulation type are usually specific to the scanner manufacturer, and it is important that the reader takes the time to familiarize themselves with the terminology used on their scanner.

X-*y*-axis modulation looks at the requirements during each rotation and decreases tube current for projections that cause less attenuation of the X-ray beam, for example, an anteroposterior projection, compared to a lateral projection, through the thorax. *Z*-axis modulation adjusts the tube current dependent upon the regional anatomy in the longitudinal axis of the patient and so, for example, would decrease the current as a scan progresses from the shoulder region into the upper thorax, but it would maintain the same current for the entire 360º rotation. The real-time modulation mode bases the tube current on the level of attenuation detected when the X-ray tube was located 180º earlier in its rotation, that is, it assumes that the path from tube to detector one way through the patient is similar to that the other way when the system has rotated through 180º (i.e., the average attenuation of a path anterior to posterior is equal to that of the path from posterior to anterior). This assumption on the average attenuation of the path is good, providing that the paths are not too far apart along the patient, that is, the pitch is not too large. All of these techniques involve the use of one or two scan projection radiographs, SPRs, to obtain the anatomical density differences necessary for the system to determine the required tube current (even the real-time systems use SPRs to determine the initial tube current). Although these SPRs are relatively low-dose scans, they still contribute to the overall patient radiation dose. However, this can usually be justified because the reduction in radiation dose provided by the use of ATCM usually exceeds the radiation dose from the SPR views.

While a primary aim of ATCM is to reduce the radiation dose to the patient, care must be taken as this is not always the case. While mA is likely to be reduced for smaller patients or less dense areas of anatomy, with larger patients or denser anatomical areas, the increase in mA that occurs might be disproportionate to the reduction of noise in the image. Using a fixed upper limit for the tube current should be considered in such circumstances. Additionally, particular patient groups require different levels of quality in an image. The quality required for diagnosis in large patients is generally lower than that required in smaller patients, due to fatty deposits causing increased organ separation in larger patients. However, if this theory is extrapolated back into pediatric patients, relatively higher image quality levels may be required compared to adults.

Correct patient positioning is also of great importance when ATCM is being used. Consider an abdomen and pelvis scan where the arms are placed in a raised position over the patient's head to reduce artifact from beam hardening and photon starvation. Now consider the same patient being positioned with their arms by their sides. To maintain image quality, an increase in mA is necessary to compensate for the extra density now within the scan plane which has been caused by the patient's arms. This will be an automatic increase when using ATCM which will in turn increase the radiation dose to the patient. Although, in certain circumstances it might not be possible for a patient to raise their arms out of the scan plane, this is an uncommon situation, and the consequence of poor positioning techniques on radiation dose and image quality must always be considered.

It is anticipated that automatic modulation systems will become ever more sophisticated alongside ever-increasing examination complexity and capability. Other dose-saving options which are currently being developed include:

- Optimum tube voltage selection and modulation
- Cardiac cycle matching—modulating the tube current so only data from the same cardiac cycle phase is obtained
- Organ sparing—reducing the tube current to very low levels (possibly even zero) during specific portions of the rotation to save dose to more radiosensitive organs, for example, the breasts

10.7 Properties of the Image Processing

The developments that have been seen over recent years in CT scanning have been made possible by significant improvements in available computing power. The number of detector elements or the speed of tube rotation could not be increased until the computing power could satisfactorily deal with the vast amount of data output from the detectors. The data not only has to be accurately detected but also processed into the required images in a time which is acceptable to the operator and/or the image reader.

As the number of detectors increases in the *z*-axis, the more the X-ray beam changes from a fan to a cone shape. This has the advantages we have already described associated with multi-slice scanning but the disadvantage that new methods and algorithms are required to reconstruct the images. A fundamental rule of the traditional reconstruction method, that is. that the incoming X-rays from adjacent rotations are parallel, is broken as the cone angle increases past that which is required for four detector rows. This means that the majority of projections through a patient, particularly at the extremes of the beam, "see" different tissues (attenuations). In traditional reconstruction techniques, it is assumed that the data collected at a particular angle of rotation of the X-ray tube and detector bank is passing through similar tissue content as data collected a subsequent rotation (or half rotation), and therefore, these two datasets can be interpolated between. Clearly, if the data from

detectors which represents paths through different tissues are utilized, the images will carry significant artifacts. This section will provide an overview of some of the techniques and parameters available for image reconstruction and viewing.

The introduction of wider and wider collimation in the *z*-axis allows imaging of complete organs in a single rotation. Current technology makes it possible to image, for example, the whole of the heart (16 cm) in a rotation time of the order of 0.2 s. This means that images can be acquired sufficiently quickly that motion artifacts do not occur. However, in radionuclide imaging, this might not prove as advantageous as it maybe in diagnostic imaging dependent on the intended use of the CT images as, for attenuation correction purposes, data acquired over the natural breathing cycle may provide a more accurate correction.

A significant benefit of volume scanning is that with no additional dose penalty, it is possible to utilize the volume of data acquired to produce multiple reconstructions. For example, the protocol can automatically include reconstructions of 0.5 mm as well as 5 mm slices, which are processed optimally for bone and/or soft tissues and also a sagittal or 3D reconstruction.

This means that the dose delivered to the patient can be used very efficiently, so long as the correct scanning parameters are selected initially. It should be noted that there is a potential to acquire "additional" data than that which is strictly required because it is so quick and easy to do so. This can take the form of including additional anatomy in the scan range, scanning the volume with too narrow an acquisition slice width or scanning the entire volume with a high level of mAs because a sub-volume requires a higher level of image quality. In this instance, consideration should be given to the option of scanning the larger volume with an adequate level of mAs and scanning the sub-volume additionally with higher factors or making use of the automatic tube current modulation system, which is particularly useful if there is a requirement to reconstruct images covering the entire scanned area, for example, in a 3D reconstruction.

It is imperative that the scanning protocols are constructed so that sufficient flexibility is available to the operator to tailor the protocol to the clinical question rather than to purely select by anatomy.

10.7.1 Image Windowing

CT images are normally constructed of a matrix of 512 × 512 pixels with each pixel being assigned one of 256 grey levels to represent the CT# of the tissue. In order that the CT# of similar tissues can be discerned by the human visual system, it is possible to adjust the minimum and maximum CT# that is displayed in the image, the so-called window width. Adjustment of the window width (minimum to maximum) and also the center point (window level) is possible in order to maximize small differences in CT#. Areas with CT# greater than the maximum will be displayed as white and those below it will be black, the CT numbers in between will be spread over the 256 available grey levels.

It is usual practice to set the window level to the average density value of the tissue being demonstrated on the image. For example, when imaging through the brain, a window level of approximately 35–40 would be used. This takes into account both the grey and white matter substance of the brain. The window width would need to be set to include the density values of relevant anatomy which falls either side of the window level. For the cerebrum, an approximate window width of 90 would be used which would allow 45 HU either side of the window level to be demonstrated. This would allow visualization of the soft tissue structures and the ventricular system but visualization of structures with CT numbers outside of this range would be limited, for example, the dense bone of the cranium would be visualized as white.

The use of a narrow window width (as in the example above) increases the contrast resolution within an image. This is important in areas such as the cerebrum where inherent contrast between the tissues is low, that is, the tissues have a similar density value. A narrow window width allows only a limited number of densities to be displayed using the grey scale so a higher number of greys are available to display a small number of densities. The converse is true when a wide window width is used to display a larger range of tissue densities, for example, in the pelvis. The same grey scale is used to display a larger number of density values so there will be less shades of grey to ascribe for the different densities leading to a reduction in contrast resolution in the resultant image. This can be overcome to an extent by the use of intravenous contrast agents which act to increase image contrast where inherent contrast between organs is low.

10.7.2 Image Reconstruction Methods

10.7.2.1 Single Slice Axial Scanning

The attenuation data obtained during the CT scan has to be reconstructed into a useful image. This typically includes a process called filtered back projection. In this process, a single value, based on the mean attenuation for the projection, is assigned to each pixel in the line of the image matrix corresponding to the projection through the patient (back projection). When all projections are computed and added together, an approximate representation of the imaged object is produced. Despite the fact that the accuracy of the representation improves as the number of projections used to construct the image increases, the spatial resolution achieved from this basic technique tends to be poor and is improved by applying filters to the data before back projecting (filtered back projection). The filter (aka kernel or algorithm) can be user selected to emphasize relevant anatomy. A sharp filter will emphasize edges and is used in, for example, sinus imaging, whereas a smooth filter will emphasize small inherent contrast variations in the image and may be utilized in liver imaging.

As always, there is a trade-off; sharp filters tend to increase the noise in the image while maintaining the spatial resolution, smoothing filters decrease both the noise and the spatial resolution. The magnitude of any noise variations is manufacturer

dependent, and clearly, if discussed in terms of a variation from a standard filter, it will depend on the noise level which is acceptable as standard. It is certainly possible to vary the noise level in the region of 10–150 HU, solely by selecting different filters for otherwise consistent acquisitions.

Many scanners are supplied with a large array of available filters, which are frequently identified by nonspecific titles. Most scanners will restrict the list of filters available to those that are suitable for use with the selected anatomy or patient age. Readers should acquaint themselves with the nomenclature adopted by their scanner manufacturer. It is suggested that the initial selection is taken with applications specialist advice, and this is followed by local investigation to determine if further optimization options are available.

It is important that the correct filter is chosen for the particular imaging task. Different filters will have varying effects on other parameters. Parameters that are likely to be affected by the choice of filter include image noise, spatial resolution, edge visibility, and CT#. It should also be noted that these parameters may be affected by the scanning kV and the level and form of any effects may be different depending on the filter and kV in question.

10.7.2.2 Helical Scanning

Basic methods of filtered back projection were developed for axial scanning; helical scanning acquires data in volume form and introduces the concept of pitch.

Helical scanning requires a modified approach to image reconstruction compared to simple axial acquisitions. Standard filtered back-projection techniques would effectively produce a helix of uniform images, as every projection is acquired at a different location along the *z*-axis of the patient. A solution to this is based on the assumption that it is possible to interpolate between data from points acquired on subsequent rotations, on either side of the required location.

If interpolation is performed between data gathered a complete rotation to either side the required location, it is referred to as a 360º interpolation. If the (reasonable) assumption is made that the attenuation remains relatively unchanged between a beam traveling north to south as when the tube and detector have rotated 180º so the beam is traveling south to north, then this location can be used for data reconstruction and the projection is referred to as "complementary." Such reconstructions are made using 180º interpolation (see Fig. 10.3).

One-hundred and eighty degree interpolation has the advantage over 360º as, because the data points are closer together, it can provide a narrower effective image slice width and is less prone to introducing partial volume artifacts into the image, particularly in regions where the anatomy is changing rapidly. It is easy to understand that, particularly for 360º interpolations, if the pitch of the helical scan is large, the data points available for the rotation are widely spaced along the patient, so care is required in pitch choice particularly for areas where the anatomy is varying rapidly or visualization of small details is required. This therefore can have a dose implication to the patient, as for fixed mAs, increasing the pitch decreases the patient dose.

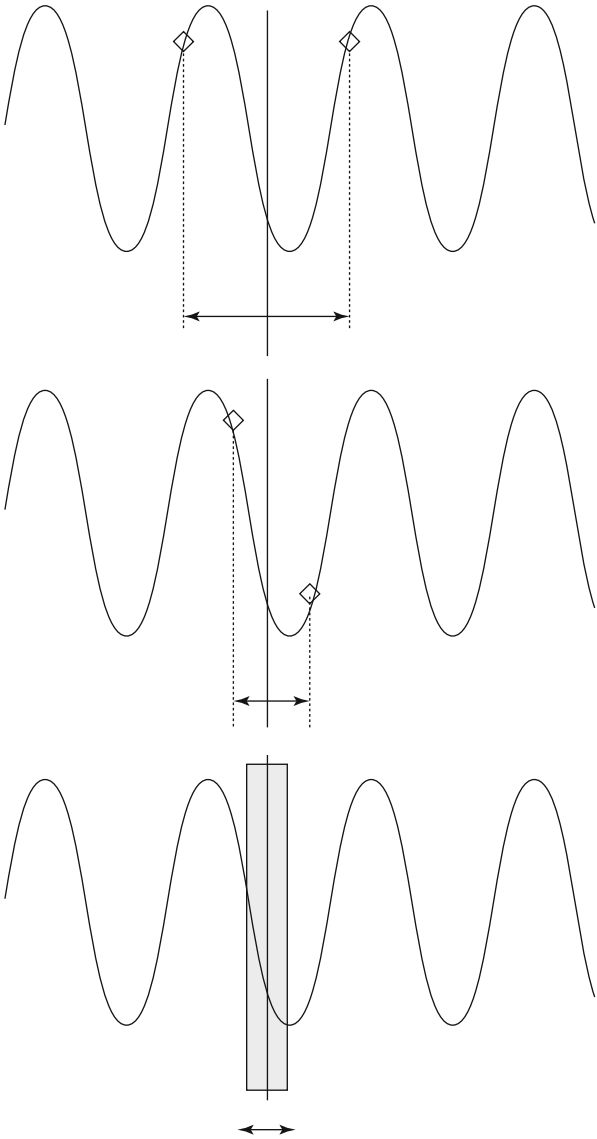

Fig. 10.3 Image reconstruction methods. (**a**) 360º interpolation. Showing slice location (*solid line*), interpolation data points (*diamond*), and effective slice width (*dotted lines*). (**b**) 180º interpolation. Showing slice location (*solid line*), interpolation data points (*diamond*), and effective slice width (*dotted lines*). (**c**) Multi-detector CT reconstruction. Showing slice location (*solid line*). Filter width (*shaded area*) shows the effective slice width

The introduction of a multi-slice CT scanning further modified the basic reconstruction techniques. The bank of detector elements in *z*-axis allows images to be reconstructed from data taken, not only from the projections acquired immediately before and following the required location, but from any of a group of detector elements which fall within a specified filter width. As the filter width is increased, more data points become available for the image reconstruction. Increasing the number of data points used will decrease the noise in the image but increase the effective slice width. Careful choice of this filter width therefore helps to balance the effective slice width and noise level.

In order to optimize the use of the data available for image reconstruction manufacturers will typically limit the choice of pitches available to the scanner operator. These pitches will be based on the selected acquisition and image slice widths, the reconstruction filter and the anatomy in question. The available pitches are defined such that there are interleaving direct and complementary projections at the isocenter available for image reconstruction. An important dose issue that arises with helical scanning is the requirement for the scanner to perform additional rotations at the beginning and end of the intended volume. This so-called over-scanning is required to provide the data points for construction of images corresponding to the beginning and end of the defined scan volume.

10.7.2.3 Multi-detector (Cone Beam) Scanning

As previously mentioned, one of the basic assumptions used in filtered back projection is that the X-ray beams from one rotation to the next are parallel, that is, there is no spreading of the beam in the *z*-axis. This holds approximately true for multi-detector systems up to approximately four slices. Once more slices are included, the X-ray beam has enough of a cone angle that projections taken from north to south pass through different anatomy to the slightly later south to north projection, that is, there are no longer direct and complimentary projections. If traditional reconstruction methods are used in this case, geometric distortions are introduced into the image, which obviously increase in severity as the cone angle of the X-ray beam increases. Hence, more sophisticated reconstruction algorithms are required in order to account for the cone angle.

One reconstruction method which incorporates the cone angle is based on an extension of simple filtered back projection methods. Instead of the back projected data accounting for a line in the image matrix, it is projected back across a cone and then all such cones are added together. This reconstruction method is very computationally intensive and so some manufacturers use methods which segment the data prior to reconstruction. These methods have the advantage of being very quick to compute. Their main "trick" is to initially compute series of image groups at intervals around the rotation, which are aligned along planes other than the traditional *x*-*y* plane, for example, planes along the direction of the cone beam. These images are generally constructed from data gathered over a partial rotation, and they interleave together (Fig. 10.4). They can then be interpolated between in order to produce images of the required orientation and thickness. The noise produced by these reconstruction algorithms is insensitive to the choice of pitch, and generally, it is effective mAs which is defined by the operator as, if the mAs was defined explicitly, the noise would increase as pitch increases due to a reduction in the number of data points in the reconstruction. This also leads to the patient dose being independent of pitch as long as the effective mAs remains constant. The patient dose equals that which would be delivered from a sequential scan of the same mAs.

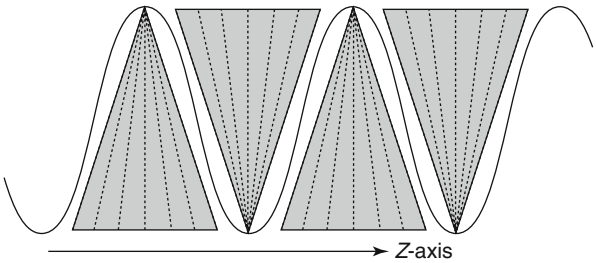

Fig. 10.4 Showing groups of images reconstructed at planes related to the cone beam, rather than in the *x-y* plane, at various projection angles, represented in the *z*-axis as a sinusoidal wave. These reconstructions together then provide sufficient information to allow images to be further reconstructed into any plane

10.7.2.4 Iterative Reconstruction

The reconstruction methods outlined so far are based on modifications of the simple filtered back projection techniques only. This method of image reconstruction is prone to producing higher and higher levels of image noise as the number of X-ray photons used is decreased and has been one of the factors slowing down progress toward low-dose imaging. Improvements in computational power has allowed introduction of a new reconstruction technique into CT imaging; it is called iterative reconstruction. This technique, although still in its infancy for CT, is helping to reduce image noise, and therefore can reduce patient doses.

The technique typically involves production of an initial estimation of the image (usually by filtered back projection for speed); this image data is then forward projected and compared to the actual data collected. These images are compared and the differences between them used to produce an error matrix. This error is then applied to the estimated image to produce a new estimate, which will be a better approximation of the actual object. This process is repeated over a number of iterations. It is important that the cycle is not repeated too much as it may introduce artifacts into the image. It is possible to incorporate information within the estimated image that models elements of the system and can minimize image noise or potential causes of artifact. These can come from many sources, for example, any nonuniformity of the beam over the detector, scattered radiation, or beam hardening.

An advantage of iterative reconstruction over pure filtered back projection is that it is able to retain high levels of spatial resolution and the number of iterative cycles is variable dependent on the imaging task so it may help to produce adequate images, for example, in cardiac studies when requirements for high temporal resolution lead to low photon counts (or very high doses).

Iterative reconstruction's demands for high computing power, with still relatively lengthy image reconstruction times, may be somewhat mitigated if the CT imaging is carried out prior to the radionuclide acquisition. This will allow time for the

images to be reconstructed before they are required for evaluation. However, this may have detrimental effects to any attenuation correction if contrast media is to be utilized for the CT image.

10.7.3 Image Artifacts

10.7.3.1 Beam Hardening

This is an effect that may be seen in areas of images where the radiation beam has passed through a region of high-attenuation tissue into lower attenuation matter. As the X-ray beam is polychromatic when it passes through tissues with high attenuation, the lower energy portion of the spectrum is preferentially absorbed, which has the effect of increasing the average energy of the beam; the beam is hardened. This distorts the ideal situation where the attenuation "seen" by the projection is directly related to the path length. This discrepancy in beam energy is translated into the image as dark regions, typically prevalent in the base of skull region, between the petrous bones or by "cupping" of the CT numbers of a region of uniform attenuation. This effect is minimized by adequate filtration of the X-ray beam to remove the softest X-rays and reconstruction algorithms will include correction factors.

10.7.3.2 Partial Volume

Partial volume artifacts are produced when the anatomy being imaged is smaller than the acquisition slice width. Each 2D picture element (pixel) forming the image represents a 3D volume element (voxel) of the patient. Different densities occurring within the voxel are averaged out and represented by a single CT number (or Hounsfield unit) in the resultant pixel. To reduce this averaging of densities and minimize partial volume effect, a narrow slice width needs to be used. When the slice width is equal to the x and y dimensions of the pixels, the image is described as being isotropic, that is, the voxels being represented have sides of equal proportion. When the voxels are isotropic, there will be no z-axis distortion, and images may be reconstructed in any plane of the acquired volume with the same level of resolution in any direction. Modern high-powered CT scanners now produce images which are isotropic, and images are typically acquired using a narrow acquisition slice width and then summed with other slice data to reduce the effect of the image noise but retain the high spatial resolution.

If a metallic implant is present in the scanned anatomy, both the beam-hardening and partial volume artifacts are magnified, and the anatomy surrounding the implant will have significantly reduced clarity. Due to the nature of the image formation, there may be some level of streaking throughout the image, although again these can be somewhat corrected for by the reconstruction algorithm.

10.7.3.3 Spiral Artifacts

Due to the image reconstruction models, projections which are not directly complementary may be utilized for reconstruction of helical scans. When imaging rapidly varying anatomy or when imaging at high pitch, this discrepancy may be translated into the reconstructed image as geometric distortions or broadening of anatomy. Considering multi-detector scanners, the axial images may be reconstructed using sequential groups of detector elements as they fall within the reconstruction filter width. This can result in a "windmill" effect as the position of the artifact varies along the *z*-axis and is most prevalent at high contrast difference boundaries. Careful choice of acquisition slice width (narrow compared to reconstruction slice width) and pitch (to maximize number of sample points within the reconstruction filter width) helps to minimize these effects.

10.7.3.4 Cone Beam Artifact

As we have already mentioned, once the cone angle of the X-ray beam passes a certain threshold size, the divergence of the beam has to be accounted for within the reconstruction algorithm. As the cone angle increases, projections reconstructed from the central elements of the detector will show very different geometry of a uniform object to those at the extremes. This will manifest as streaks in the images but will be minimized by corrections within the reconstruction algorithm.

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Part V
Imaging Principles

Chapter 11
Cross-Sectional Anatomy in Multi-Planar Imaging

John M. Delieu and Joanne Sil

A necessary requirement for a truly academic understanding of any structural imaging modality is a thorough underpinning of basic human anatomy. With this in mind, the following is hopefully a foundation to equip you into a deeper understanding of the complexities of the structural organization of body structures. Underpinning this is the need to know the surface markings which is fundamental for the correct positioning and location of organs of the patient undergoing an imaging procedure.

The forthcoming sections are laid out in a sequential manner, from the initial anatomical position and the basic embryological development to the surface anatomy and its underlying structures.

11.1 A General Plan of Surface Anatomy

Taking the anatomical position (see Fig. 11.1a, b) as a starting point for any descriptive portrayal of the human body is of first importance. Once this is imprinted on the practitioners mind, the rest follows through in a logical and systematic process. These figures will be referred to throughout this chapter, and the reader is advised to this fact. Notice that the line *A* goes through the sagittal suture (see Fig. 11.2) as the sagittal plane, therefore dividing the body into a left and a right side.

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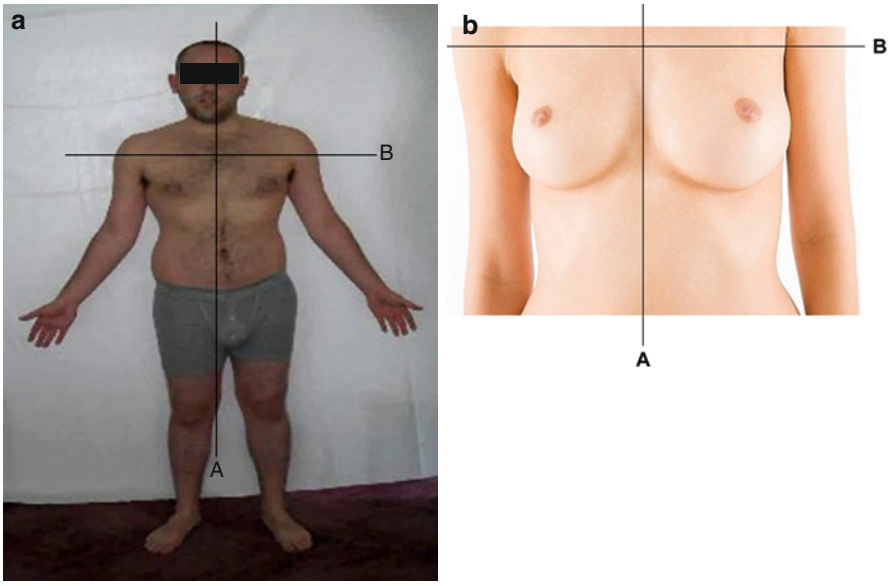

Fig. 11.1 (**a**) The anatomical position 2. (**b**) Showing female breast in the anatomical position

The line *B* shown in Fig. 11.1a, b is referred to as a transverse or axial plane, and sometimes as horizontal. For purposes of consistency, the axial plane will be used when referring to the CT images. Indeed, all the CT levels are viewed in the axial plane and looking from below.

11.2 Planes Based on the Basic Principle of the Anatomical Position and the Sutures of the Skull

It is from this basic concept, which is of utmost importance, that the terms sagittal, coronal, and transverse/axial are derived. They are used in all descriptive purposes.

Firstly, let us look at the top of the skull (calvaria) as the reference point; then if one looks down onto the surface of the skull from above, prominent suture lines are observed (see Fig. 11.2). The two major ones, used for descriptive purposes, are the sagittal and coronal sutures. These in namesake are the bases for the definition of the sagittal and coronal planes. A line passing through the sagittal plane neatly bisects the body into a left and right side. At right angles to this plane is the coronal plane; likewise, this divides the body into an anterior and posterior.

11.3 Lines

There are other reference sites of importance, in particular lines which underpin planes. Drawn imaged lines can often help in identifying structures immediately

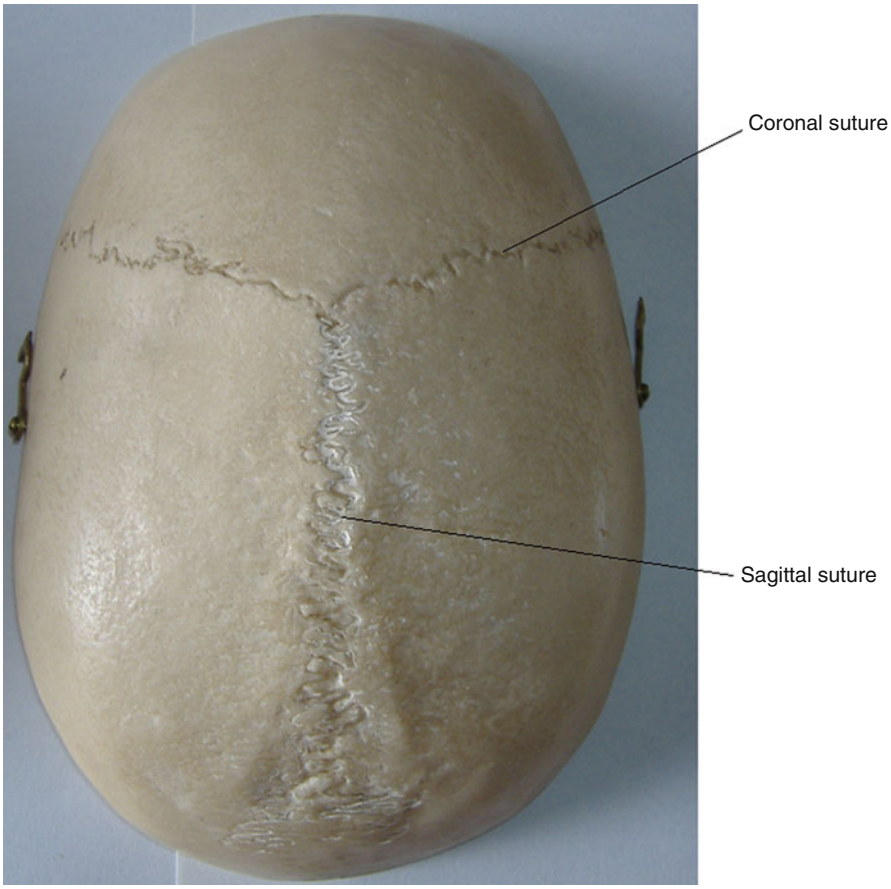

Fig. 11.2 Calvaria showing suture lines

beneath the surface. Take a line drawn in the midclavicular; this will be equivalent to a parasagittal plane. One can also use these as lines of reference when dealing with surface anatomy, i.e., surface markings of the pleura. This is dealt with more fully later on in this chapter.

11.4 Basic Embryology: How the Anatomical Position Arises from the Rotations of Limbs at the Embryological Stage

In the womb, huge morphological changes are continually underway in a timely order. The early stages of the embryo resemble that of a "tadpole," having gills and a tail. As the organogenesis of the embryo progresses, there is active differentiation of an organ or tissue. This occurs between the 4th and 8th week of gestation [1], and exposure to teratogens during this period may cause congenital defects as also does radiation (see Sect. 9.3). The embryo takes on a characteristic C-shaped curvature

(in the neonate this is called the primary curvature). Upper limb buds are recognizable by day 26 or 27 with the lower limb buds following shortly after. Early in the 7th week, the limbs extend ventrally, and at this point the developing upper and lower limbs rotate in opposite directions. The upper limbs rotate laterally through 90° on their longitudinal axes, this causing the future elbows to point backwards, with extensor muscles lying lateral and posterior. The converse is true of the lower limb whereby there is 90° medial rotation; hence, the future knees face forwards, with extensor muscles lying anterior.

This puts the embryo into its primary "anatomical position," but remembering it is still C-shaped. The next stage, to a bipedal anatomical stance, is the development of secondary curvature points which occur in a timely manner after birth. The first to appear is the cervical curvature when the baby is 6–8 weeks old, with further development occurring when crawling around. Babies are inquisitive, and to explore their environment, they need to be able to raise their heads. It is no good looking in a downward direction (position of eyes in humans is different from other animals). The next and final stage is when the skeletal and muscular support is adequate enough for the toddler to gain a standing gait and to be able to explore the surrounding environment further. This then is the completed anatomical position, portrayed in Fig. 11.1a.

11.5 Relationship of Regional Anatomy to Surface Markings

There is no better place to start than from the top of the body and moving downward in a systematic order: from head to toe. The regions we will concentrate on in this section are (1) head and neck, (2) thoracic, (3) abdo-pelvic.

11.5.1 Head and Neck Region

11.5.1.1 Surface Markings

One can revise the important landmarks of the skull on oneself. The important surface markings to commence with are on the facial surface; first is the glabella. This is defined as the highest point between the left and right brows of the eyes. There is then a point of inward reflection called the nasion. Moving downward in an inferior direction is the bridge of the nose. On the back of the head is the inion; lying midline it is a prominence which is sometimes called the external occipital protuberance. A major reference point is the mastoid process, a point of attachment for the sternocleidomastoid muscle.

Moving onto the neck region, there are two important territorial areas defined as the anterior and posterior triangles. The major player in this is the sternocleidomastoid muscle which divides the anterior and lateral aspects of the neck. The region in front is defined as the anterior triangle containing important structures such as the carotid

sheath, hyoid bone, and thyroid cartilage. Behind the sternocleidomastoid muscle, the posterior triangle is evident. Here important structures are contained: brachial plexus, accessory nerve, to name a few.

11.5.1.2 Vertebral Levels

The hyoid bone lies at the level of C3, whereas the lower border of the cricoid is at C6. This is also where commencement of the trachea occurs. The bifurcation of the common carotid artery occurs at C4. One other landmark to mention is the suprasternal notch which lies between the medial ends of the clavicles, which is at the level corresponding to the lower border of body of T2.

11.5.1.3 Cross-Sectional Images

The following CT images will be referred to within the main text. It must be emphasized that all the labelled images referred to, are in the axial plane, i.e. with one looking in a cephalic direction. Therefore, please note that in Fig. 11.3, label (a) refers to the right lobe of the thyroid and conversely (g) is the right pectoralis major muscle.

11.5.1.4 Underlying Structures

Contained within the skull is the brain, together with its meningeal coverings: pia, arachnoid, and dura mater. Figures 11.4, 11.5, 11.6, and 11.7 show various axial slices. Notice on Figs. 11.6 and 11.7 the intense white calcification, which is often helpful in identifying and orientation of anatomical markers within the skull. The pineal gland and choroid plexus often show up as calcified structures. Figure 11.6 shows a calcified pineal gland (labeled e). Evident in Fig. 11.7 is the calcified choroid plexus (labeled b), which is contained within the lateral ventricles.

The important neck structures are the seven cervical vertebrae and their associated muscles which all lie posteriorly. Anteriorly is the pharynx and esophagus, and immediately in front is positioned the larynx and trachea. The thyroid gland lies on

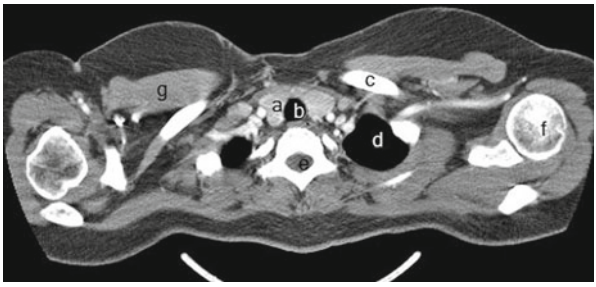

Fig. 11.3 CT image showing the anatomical structures at the level of the 2nd thoracic vertebra: *a* thyroid, *b* trachea, *c* left clavicle, *d* apex left lung, *e* spinal canal containing spinal cord, *f* left humeral head, *g* right pectoralis major muscle

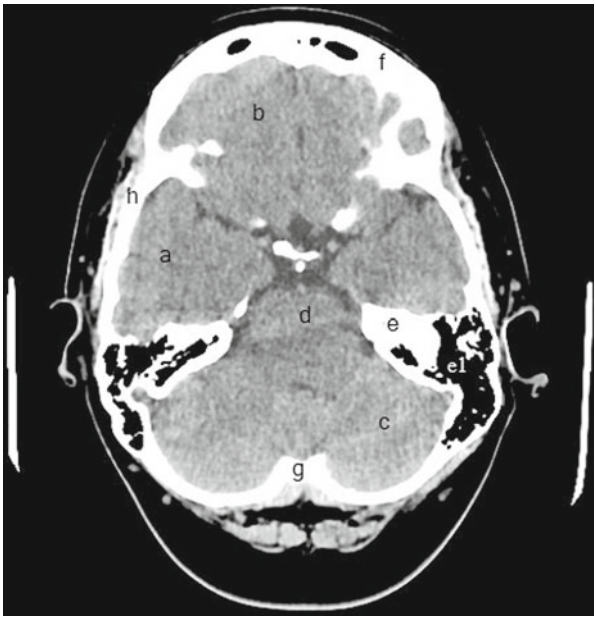

Fig. 11.4 CT image of a section through the head showing structures found within the regions of the anterior, middle, and posterior cranial fossae: *a* right temporal lobe, *b* right frontal lobe, *c* left cerebellar hemisphere, *d* medulla oblongata, *e* petrous part of the left temporal bone containing mastoid air cells (*e1*), *f* frontal bone, *g* internal occipital protuberance, *h* squamous part of right temporal bone

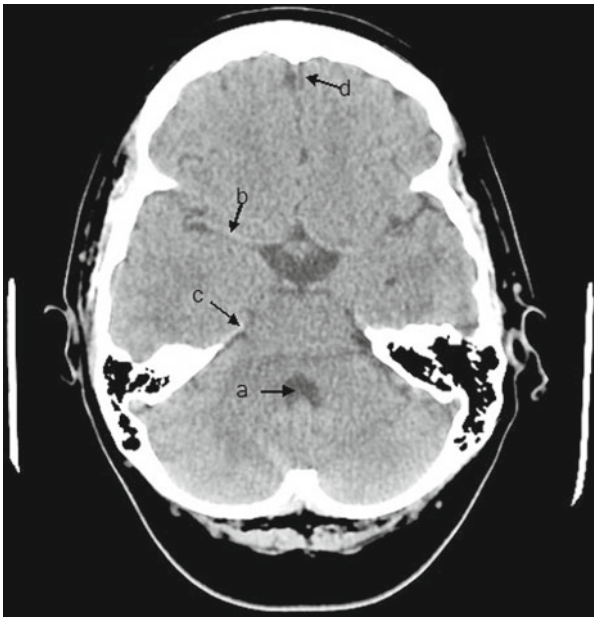

Fig. 11.5 CT image of a section through the head showing structures found at the level of the 4th ventricle: *a* 4th ventricle, *b* right middle cerebral artery (part of circle of Willis), *c* tentorium cerebella, *d* falx cerebri

either side of the lower extremity of the larynx and trachea (these can be seen on the image of the T2 axial plane in Fig. 11.3, labeled a). Indeed, the gland has a left and right lobe interconnected by the isthmus, covering the second and third tracheal rings. Parathyroid glands are embedded on the posterior surface. These vary in number but usually around four.

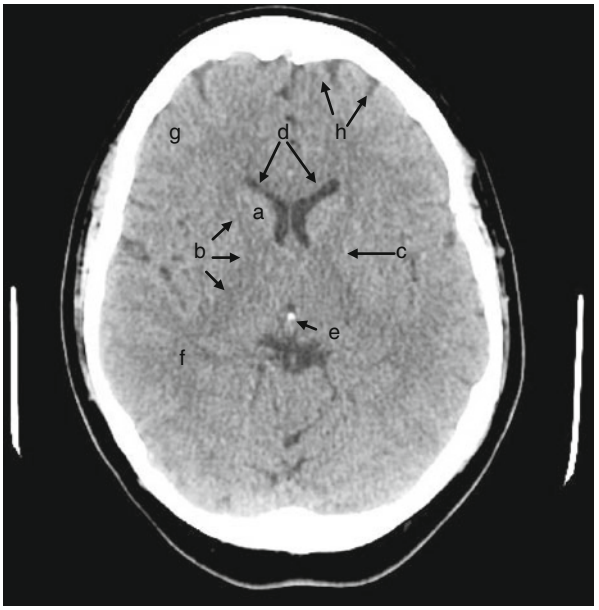

Fig. 11.6 CT image of a section through the head showing structures found at the level of the pineal gland: *a* right caudate nucleus, *b* right internal capsule, *c* lentiform nucleus (consisting of globus pallidus and putamen), *d* anterior horns of the left and right lateral ventricles, *e* calcified pineal gland, *f* white matter, *g* grey matter, *h* sulci

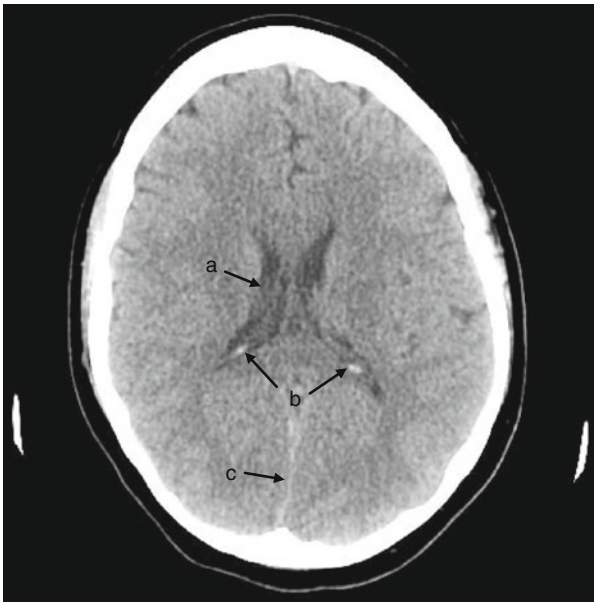

Fig. 11.7 CT image of a section through the head showing structures found at the level of the lateral ventricles: *a* body of right lateral ventricle, *b* posterior horns of left and right lateral ventricles containing calcified choroid plexus, *c* falx cerebri

With a finger placed between the sternocleidomastoid muscle and the upper part of the thyroid cartilage, one can feel one's own carotid pulse. This is the common carotid artery which is within its sheath of fascia together with the internal jugular vein and the vagus nerve (the 10th cranial nerve).

11.5.2 Thoracic Region

11.5.2.1 Surface Anatomy

The thorax consists of the 12 pairs of ribs, articulating from behind with the thoracic vertebrae and in front with the sternum. The lower boundary of the thorax is called the costal margin, formed by the cartilages of the 7th, 8th, 9th, and 10th ribs. A most important surface landmark is that of the sternal angle (sometimes referred to as the angle of Louis). This can be visualized by looking at the line B in Fig. 11.1a, b. Indeed, this line bisects the angle. The reason that this point is of utmost importance is that this is where the first of the palpable ribs is felt (2nd costal cartilage). Therefore, surface markings of the heart, great vessels, and lung can be traced out onto the thorax with confidence. The nipple lies in the 4th intercostal space about 10 cm from the midline, and the apex beat of the heart can be auscultated in the 5th left intercostal space, 9 cm from midline. This is generally the case in the male; however, the position of the nipple in the female is variable and extreme. The surface markings of the breast extend from the second to the sixth ribs and overlie the pectoralis major muscle.

The surface markings of the lungs have the apex plus cervical pleura, projecting into the neck, 2.5 cm above medial end of clavicle. The anterior border of the right lung begins behind the sternoclavicular joint and runs downward behind the sternal angle to the xiphisternal joint, whereas the anterior border of the left lung follows a similar course except at the level of the 4th costal cartilage; it deviates laterally so forming the cardiac notch.

The posterior border of the lung commences from the spinous process of the 7th cervical vertebra to the level of the 10th thoracic vertebra. There is an oblique fissure which divides the lungs into upper and lower lobes. This fissure follows the course of the 6th rib. The right lung has an additional fissure, which is represented by drawing a line along the right 4th costal cartilage to meet the oblique fissure. This then gives the lung an additional lobe.

11.5.2.2 Vertebral Levels

The upper border of the manubrial sterni lies at the level of the 2nd thoracic vertebra. The next important landmark is the sternal angle; this lies at the level between the 4th and 5th thoracic vertebrae. At the level of the 9th thoracic vertebra is the junction of the xiphisternum with the body of the sternum (see Fig. 11.8). The lowest part of the costal margin lies at the level of 3rd lumbar vertebra.

11.5.2.3 Reflections of Pleura

The lower border of the pleura follows a curved line which crosses the 8th rib in the midclavicular line, the 10th rib in the midaxillary line. It then reaches the 12th rib

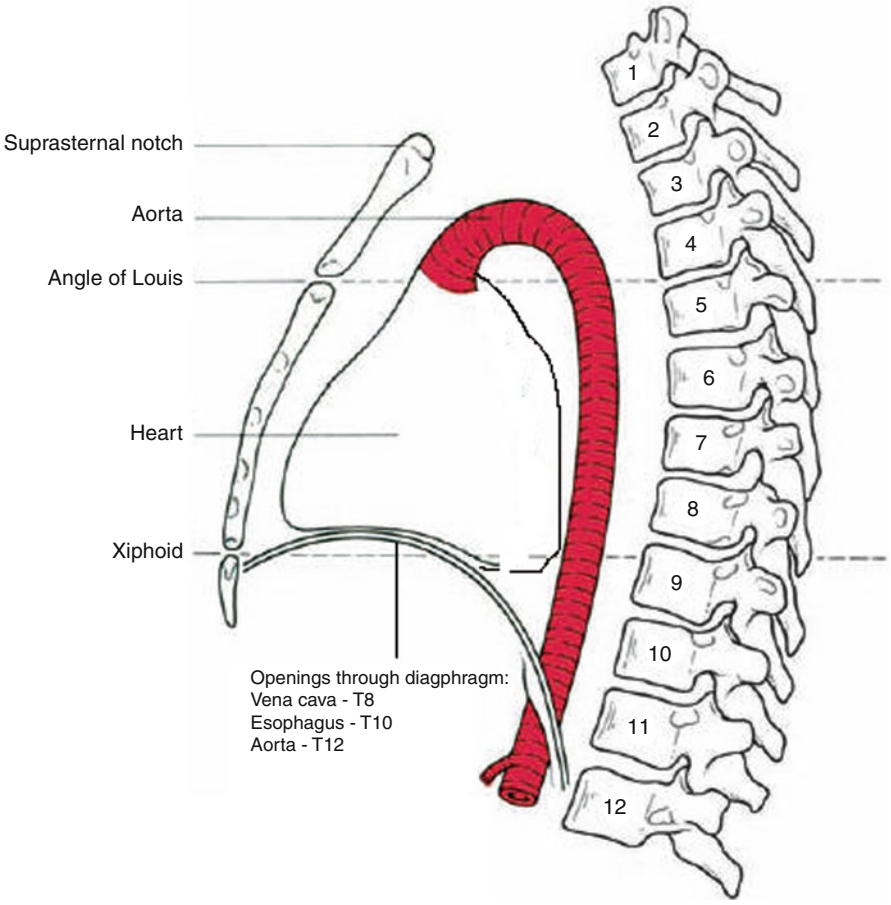

Fig. 11.8 Showing a sagittal section through the mediastinum

which is adjacent to the vertebral column. Note that the lungs cross at the 6th, 8th, and 10th; hence, the distance between the two borders corresponds to the costodiaphragmatic recess.

11.5.2.4 Mediastinum

This consists of the superior and inferior regions. The superior lies above the plane that bisects the sternal angle and the intervertebral disc between T4 and T5. This contains the aortic arch (clearly seen in Fig. 11.9, labeled (b)), together with its main branches, superior vena cava and brachiocephalic veins, trachea, and esophagus. The inferior mediastinum is subdivided into the anterior mediastinum, in front of the pericardium (visceral membrane of the heart), the middle mediastinum, occupied by the pericardium and its contents (heart and vessels), and the posterior mediastinum which lies behind the pericardium.

11.5.2.5 The Diaphragm

This is the muscular division between the structures that lie in the thoracic cavity and those that lie within the abdominal cavity. It is dome shaped and attaches to the costal margin and to the anterolateral surface of the vertebral column, as far down as L3 on the right and L2 on the left. Blood vessels, nerves, and lymphatics have to go through the diaphragm at different vertebral levels. A way of remembering what goes through at a particular level is to remember the following:

> COA 8, 10, 12. Cava (together with the phrenic nerve) goes through at the 8th thoracic vertebra, esophagus (together with branches of the left gastric artery and the vagi) at the 10th thoracic vertebra, and the aorta (together with the thoracic duct and azygos vein) at the 12th thoracic vertebra. These last two structures can clearly be seen in Fig. 11.10 at the level of the 10th thoracic vertebra (labeled e and c, respectively).

11.5.2.6 Cross-Sectional Images/Dissections

Figures 11.9, 11.10, 11.11, and 11.12 show a range of thoracic levels with the important structures for each level clearly defined.

11.5.2.7 Underlying Structures

The major underlying structures within the thoracic cavity consist of the heart and lungs. The heart is contained within its fibrous pericardium and serous membranes and is lying between the two lungs. These in turn are enveloped in pleura. Thymic residue and aberrant inferior parathyroid glands may be evident. The latter may be situated in front of the trachea and may even track into the superior mediastinum in company with the thymic tissue.

11.5.3 Abdo-pelvic Region

This refers to everything that is contained beneath the diaphragm. One must not forget that a lot of the abdominal viscera projects into the pelvic cavity.

11.5.3.1 Surface Markings

Referring to Fig. 11.13, one can see that it is divided into nine regions by two vertical lines and two horizontal lines. Each of these vertical lines passes through a midpoint between the superior anterior iliac spine and the symphysis pubis (inguinal ligament). The upper horizontal line (equates to the subcostal plane) connects

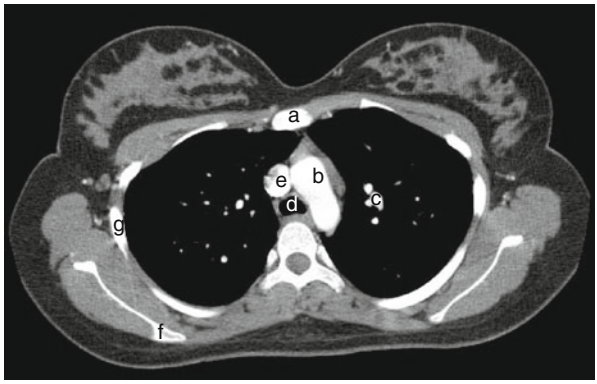

Fig. 11.9 CT image showing the anatomical structures at the level of the 4th thoracic vertebrae: *a* sternum, *b* aortic arch, *c* pulmonary vessels, *d* bifurcation of trachea, *e* superior vena cava, *f* right scapula, *g* rib

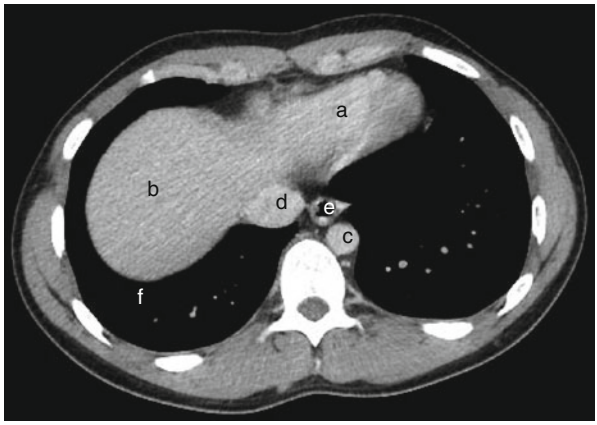

Fig. 11.10 CT image showing the anatomical structures at the level of the 10th thoracic vertebrae: *a* left lobe of liver, *b* right lobe of liver, *c* descending thoracic aorta, *d* inferior vena cava, *e* esophagus, *f* base of right lung

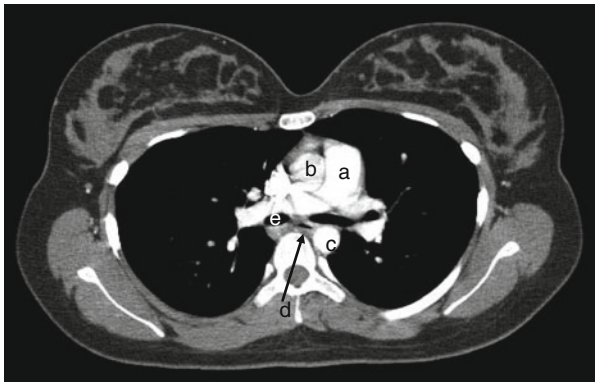

Fig. 11.11 CT image showing the anatomical structures at the level of the 6th thoracic vertebrae: *a* pulmonary trunk, *b* ascending thoracic aorta, *c* descending thoracic aorta, *d* esophagus, *e* right main bronchus

the lowest point of the costal margin on either side (inferior margin of the 10th costal cartilage which lies opposite the 3rd lumbar vertebra). The lowest horizontal line (equates to the transtubercular plane) connects the tubercles of the iliac crests. It lies opposite the 5th lumbar vertebra.

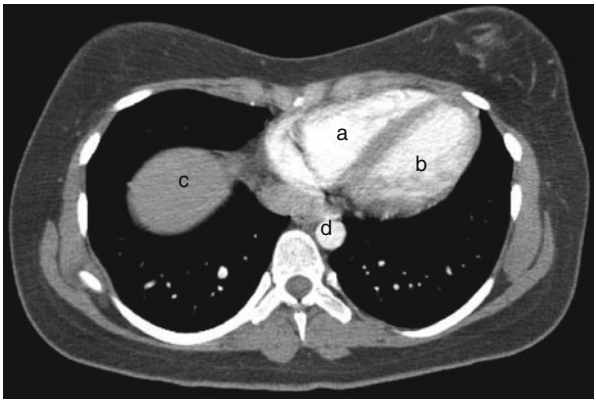

Fig. 11.12 CT image showing the anatomical structures at the level of the 9th thoracic vertebrae: *a* right ventricle, *b* left ventricle, *c* right lobe of liver, *d* descending thoracic aorta

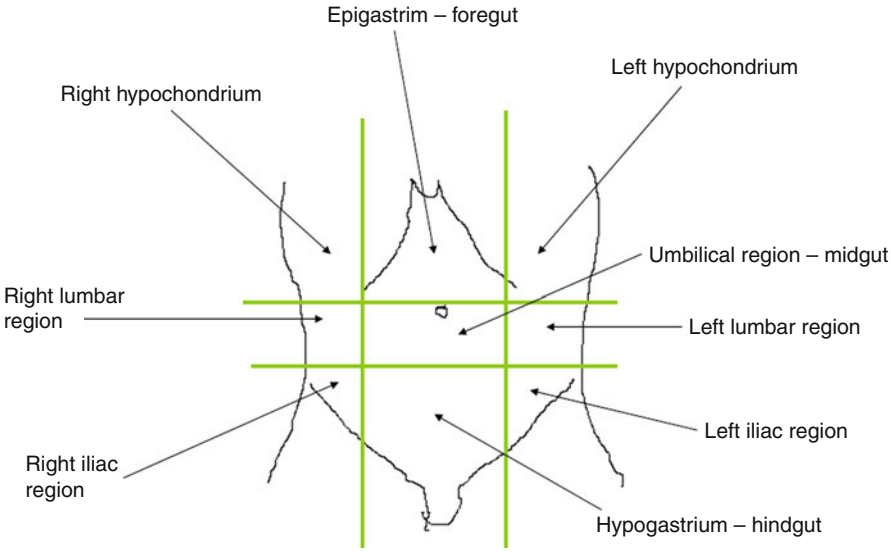

Fig. 11.13 Showing the various regions of the abdominal surface

Another plane to bring to attention is the transpyloric plane which lies halfway between the suprasternal notch and the pubis (approximately a hand's breadth below the xiphoid). This plane passes through the pylorus, neck of the pancreas, duodenojejunal flexure, fundus of the gall bladder, tip of the 9th costal cartilage, spleen, and the hila of the kidneys. This plane lies at the level of the 1st lumbar vertebra. Incidentally, the spinal cord terminates at this level. This L1 level can be seen in Fig. 11.14, showing the structures that have been noted above.

The pelvic cavity is continuous with the abdominal cavity above. The anterior wall is formed by the musculature extensions and aponeurosis of the anterior abdominal wall, together with the symphysis pubis and pubic bone. Posterior is the sacrum, coccyx, and ilium.

11.5.3.2 Vertebral Levels

The xiphoid process is the lowest point of the sternum and lies at the vertebral level of T9. The next landmark down is that of the 1st lumbar vertebra where the transpyloric plane can be found. This has already been described in some detail above. At the level of the 3rd lumbar vertebra lies the subcostal plane. Again this has been described previously. At the vertebral level of L4 is the highest point of the iliac crests. They correspond to the level of the aortic bifurcation into left and right iliac arteries. It is also a very useful landmark for the procedure of a lumbar puncture. The umbilicus lies at the junction of the 3rd and 4th lumbar vertebrae, but it is not a consistent landmark, i.e., in obesity and in pregnancy.

11.5.3.3 Reflections of Peritoneum

Those parts of the gut that are most mobile in the processing of food are contained in the abdomen, surrounded by a serous membrane, the peritoneum, to allow friction-free movement. It can be thought of as a large sac which is reflected onto the abdominal viscera (organs, e.g., stomach). You can demonstrate this yourself by pushing your hand into a large clear polythene bag. Your hand represents an organ, and this can clearly be seen through the bag. The lining immediately in contact with your hand is the visceral layer. The outside of the bag represents the parietal layer (this represents the walls of the abdomen). Your hand only represents one organ; therefore, get a colleague to place their hands within the bag, meeting up with yours. These other hands represent more organs, but notice that the hands are not in direct contact with each other; there is a double layer of plastic between them. Think of this as the serous membrane.

During the development of the abdominal viscera, some of these gradually extend into the posterior wall of the sac and become enveloped by peritoneum. They remain connected to the posterior wall of the abdomen by mesenteries (double layer of peritoneum). These convey arteries, veins, nerves, and lymphatics to the various organs. Referring back to our polythene bag model, your forearms would represent the mesenteries.

Other viscera project only slightly from the posterior abdominal wall, and only their anterior surfaces are covered by peritoneum. These organs would then be described as being retroperitoneal. Again we can use our plastic bag model by laying the palm of our hand against the bag, the palm being surfaced by the bag but the rest of the hand not so.

11.5.3.4 Cross-Sectional Images/Dissections

A selection of CT images of axial sections within the abdo-pelvic region is shown in Figs. 11.14, 11.15, and 11.16.

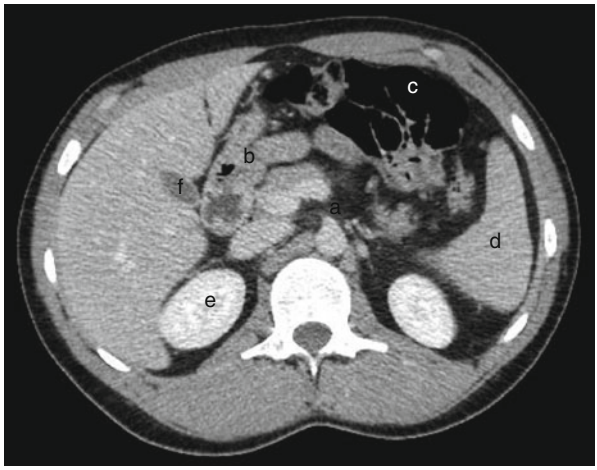

Fig. 11.14 CT image showing the anatomical structures at the level of the 1st lumbar vertebrae: *a* origin of superior mesenteric artery, *b* duodenum, *c* splenic flexure colon, *d* spleen, *e* right kidney, *f* gall bladder

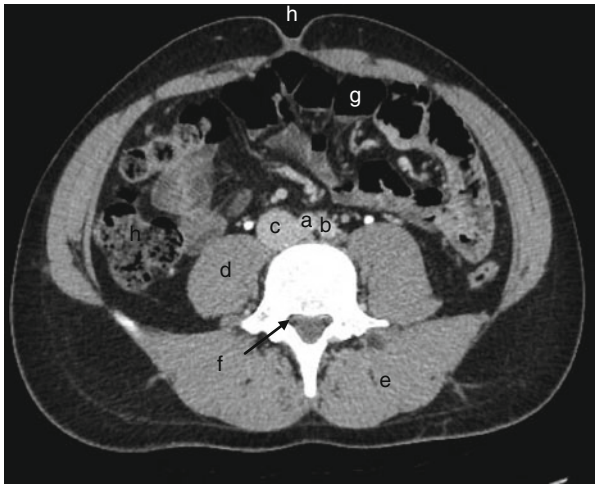

Fig. 11.15 CT image showing the anatomical structures at the level of the 4th lumbar vertebrae: *a* right common iliac artery, *b* left common iliac artery, *c* inferior vena cava, *d* right psoas muscle, *e* left erector spinae muscle, *f* right lumbar nerve root, *g* transverse colon, *h* umbilicus

11.5.3.5 Underlying Structures

The upper part of the stomach lies beneath the left dome of the diaphragm with the esophagus entering it at the 11th thoracic vertebra (remember that the esophagus goes through the diaphragm at T10). The stomach leads to the duodenum via the pylorus, lying on the transpyloric plane (L1). The duodenum is continuous with the jejunum at the level of the 2nd lumbar vertebra.

The liver lies immediately under the diaphragm, with its lower border extending along a line from the tip of the right 10th rib to the left 5th intercostal space in the midclavicular line. The upper border follows a line that passes through the 5th intercostal space on either side. The gallbladder lies at the tip of the 9th costal cartilage.

As already mentioned, the neck of the pancreas lies at L1, with the head of the pancreas lying to the right and in a downward direction. The body and tail are

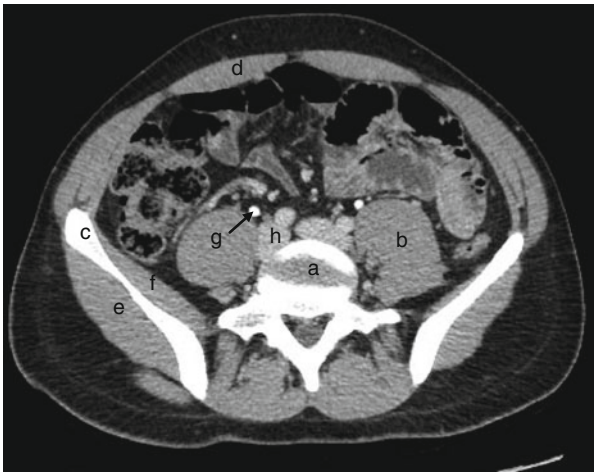

Fig. 11.16 CT image showing the anatomical structures at the level of the 5th lumbar vertebrae: *a* L5/S1 intervertebral disc, *b* left psoas muscle, *c* right iliac bone, *d* right rectus abdominis muscle, *e* right gluteal muscles (minimus, medius, and maximus), *f* right iliacus muscle, *g* right ureter, *h* right common iliac vein

directed upwards and to the left. The spleen is under cover of the left 9th, 10th, and 11th ribs, against its posterior surface.

The kidneys are two in number with the right lower than the left due to the right lobe of the liver immediately above and in front. Anteriorly the hilum lies on the transpyloric plane, four fingerbreadths from the midline. Posteriorly the upper pole of the kidney lies adjacent to the 12th rib. The left kidney lies about 2.5 cm higher than the right. The kidneys are described as being retroperitoneal.

One must make mention of the male and female structures. In the male the bladder lies anteriorly. At its base lies the prostate gland, with the seminal vesicles behind this. Lying posteriorly to the bladder and prostate gland are the rectum and lower part of the sigmoid colon. Within the female the vagina and cervix lie between the bladder and rectum. The uterus lies on the superior surface of the bladder, with the uterine tubes projecting laterally.

11.6 Common Reference Points/Planes

In the previous sections, many references have been made to vertebral levels and the importance of these to underlying structures. These can only be approximates as one must appreciate the individual variation with build and statue – e.g., C3, hyoid bone; C6, commencement of trachea; T4/T5, sternal angle; and T9, xiphisternum. There may also be differences in vertebral levels, depending on which authored text is being referred to, e.g., T7 – inferior angle of scapula in the text of Snell [2] and Gray [3], but it is T8 with Ellis [4] and Dean [5]. With the transpyloric plane, L1, and plane of iliac crests, L4, these can only be approximates, depending on the position of the individual, i.e., lying or standing. The reader is advised to refer to Table 11.1 where many of these reference points/planes/vertebral levels are summarized.

Table 11.1 Showing at what vertebral level a particular structure is located

Structure	Vertebral level
Hyoid bone	3rd cervical vertebrae
Bifurcation of the common carotid artery	4th cervical vertebrae
Notch of the thyroid cartilage	5th cervical vertebrae
Cricoid cartilage	6th cervical vertebrae
Upper limit of the oesophagus	6th cervical vertebrae
Isthmus of the thyroid gland	7th cervical vertebrae
Upper border of the manubrium sterni	2nd thoracic vertebrae
Upper border of scapula	
Upper border of arch of the aorta	3rd thoracic vertebrae
End of aortic arch	
Bifurcation of trachea	4th thoracic vertebra
Oesophageal hiatus in diaphragm	10th thoracic vertebra
Aortic hiatus in diaphragm	
Origin of coeliac axis from abdominal aorta	12th thoracic vertebra
Kidneys	Between 12th thoracic and 3rd lumbar vertebrae
Origin of superior mesenteric artery from abdominal aorta	1st lumbar vertebra
Right crus of diaphragm attaches	1st to 3rd lumbar vertebrae
Left crus of diaphragm attaches	1st to 2nd lumbar vertebrae
Origin of inferior mesenteric artery from abdominal aorta	3rd lumbar vertebra
Bifurcation of aorta	4th lumbar vertebra
Left and right iliac veins join to form inferior vena cava	Lower border 4th lumbar vertebra

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Chapter 12
Radiological Contrast Media and Injector Systems

Jonathan Priestley and Joanne Sil

12.1 Contrast Medium

12.1.1 Introduction

Since Roentgen discovered X-rays in 1895, they have remained an integral part of medical practice. Medical demands placed on X-ray imaging techniques have necessitated the continual development of the imaging technology; alongside this the need for contrast medium as an integral component of many X-ray procedures has emerged. In 2004, it was estimated that approximately 60 million doses of contrast medium are administered worldwide each year [1].

This chapter will focus on the use of contrast medium (agent) within CT. Specifically, it will attempt to describe the use of intravenous contrast medium and the use of pressure injector delivery systems. Within CT, a positive contrast medium has a wide range of clinical applications. In particular, it allows exquisite detail to be obtained of vascular systems throughout the body along with the demonstration of hypervascular organs such as the liver or kidneys.

Contrast medium is used to assist in the diagnostic accuracy of a variety of radiological tests. Contrast can be defined as the perceived difference between two adjacent structures. Within radiography, contrast can be defined as the difference in optical

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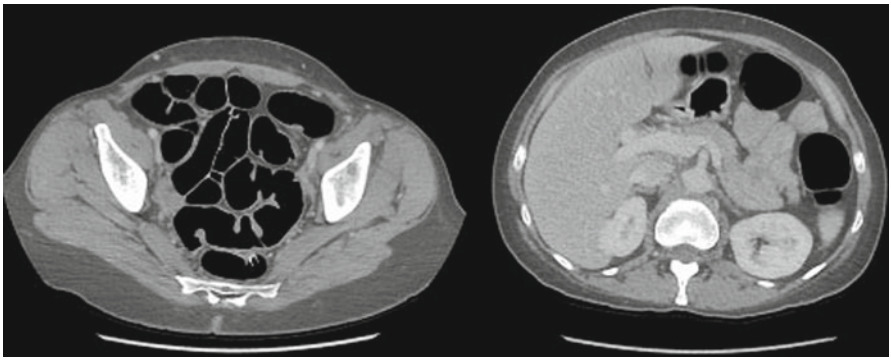

Fig. 12.1 Demonstrating the use of carbon dioxide as part of a CT colonography examination

density in a radiograph that results from a difference in radiolucency or penetrability of the subject [2]. When natural contrast exists between two structures within the body, it can be described as having good inherent contrast resolution. Examples would be between the dense cerebral tissue of the brain and the low-density cerebrospinal fluid found with the ventricles, or within the variety of tissue densities demonstrated within the thoracic cavity when imaging the lung tissue. It is usually the case that differentiation of those adjacent structures can be realized without the need for contrast medium. The primary aim of contrast medium is to increase the differentiation between adjacent structures by increasing the difference in attenuation of the X-rays by an organ or tissue. Soft tissue throughout the body does not sufficiently absorb X-rays [3] and as such inherent contrast between soft tissues can be low. Contrast agents are also useful to help characterize pathology and to demonstrate vascular structures.

Generally speaking, contrast medium is positive – that is, it will produce a higher attenuation value than surrounding tissue. It can also be negative – this will produce a lower attenuation value than surrounding tissue. Intravenous contrast medium is one example of a positive contrast medium. Carbon dioxide or air is an example of a negative contrast agent; this can be introduced into the rectum during a CT colonoscopy examination (see Fig. 12.1). Another commonly used negative contrast agent is water which is often used for CT studies where imaging of the esophagus and stomach is important. The use of negative contrast agents in such examinations allows good demonstration of the gastric mucosa which is often obscured due to artifact from a positive contrast agent.

Figure 12.1 demonstrates the use of negative contrast agent (in this case carbon dioxide as part of a CT colonography examination) which has been used to insufflate the colon. This acts to distend the colon and eliminate haustral folds which could obscure pathology such as a small polyp and also gives the colon a negative contrast allowing excellent visualization of the bowel wall. Note that the image to the right demonstrates the use of positive contrast in the form of intravenous contrast media which can clearly be seen in the kidneys, aorta, superior mesenteric artery, and hepatic portal vein.

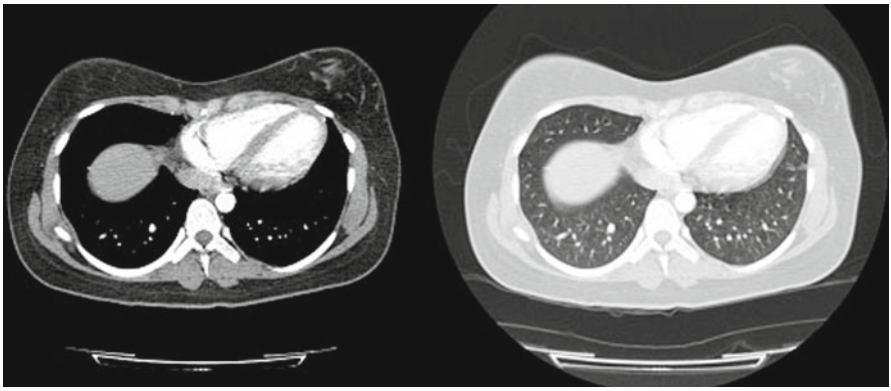

Fig. 12.2 Demonstrating the use of positive contrast in the form of intravenous contrast medium

Figure 12.2 demonstrates the use of positive contrast in the form of intravenous contrast medium. The images form part of a CT pulmonary angiographic study, and as such they have been acquired in arterial phase; the images clearly demonstrate the pulmonary vessels and chambers of the heart. These two images also demonstrate the exceptional inherent contrast properties of the thorax. The lungs have low attenuation as they contain air which acts as a natural negative contrast agent.

Intravenous contrast media are water soluble and iodine based with three iodine atoms being attached to a benzene ring. They can occur as monomers, where the molecules consist of one tri-iodinated benzene ring, or dimers, where the molecules consist of two tri-iodinated benzene rings.

Contrast agents can be divided into ionic or nonionic depending on whether the molecules dissociate in water or not. Ionic contrast agents dissociate into positive and negative ions; no dissociation occurs with nonionic contrast agents.

Contrast can be further divided into high osmolar contrast media (HOCM), low osmolar contrast media (LOCM), and iso-osmolar contrast media (IOCM). HOCM have five to eight times the osmolality of plasma, LOCM have two to three times the osmolality of serum, and IOCM have the same osmolality as blood and cerebrospinal fluid [4].

When a contrast agent is introduced into a vessel, it draws water by osmosis from the surrounding tissues; hence, it exerts osmotic pressure. The osmolality of a contrast agent is dependent upon its dissolved particle concentration. Consequently, as ionic contrast agents dissociate in water, they tend to have a higher osmolality (more particles) than nonionic contrast agents. Generally speaking, HOCM are not tolerated as well as contrast media with an osmolality closer to that of body fluids, for example, blood and cerebrospinal fluid. Higher osmolality is responsible for symptoms such as heat, discomfort, and pain [5].

Contrast media can vary greatly in viscosity and in osmolality. Early contrast media had high osmolalities, and subsequently their use led to more adverse reactions than current media. In further attempts to reduce osmolality, iso-osmolar media were developed, which was only possible with a corresponding increase in viscosity.

Table 12.1 Some of the most common CT contrast media in use today

Trade name	Manufacturer	Pharmaceutical name	Concentrations	Additional information
Visipaque	GE Healthcare	Iodixanol	320, 270	Iso-osmolar, nonionic
Niopam	Bracco	Iopamidol	150, 200, 300, 340, 370	Low-osmolar, nonionic
Omnipaque	GE Healthcare	Iohexol	140, 180, 210, 240, 300, 350	Low-osmolar, nonionic
Ultravist	Bayer	Iopromide	150, 240, 300, 370	Low-osmolar, nonionic
Optiray	Covidien	Ioversol	240, 300, 320, 350	Low-osmolar, nonionic

There are many different types of contrast medium, and although they are similar in their applications, their core properties vary from manufacturer to manufacturer. Table 12.1 shows some of the most common CT contrast media in use today along with various specifications. It is important to note that not all of these contrast media are suitable for intrathecal use.

12.1.2 Pharmacology and Good Practice in Contrast Administration

As previously stated, all intravenous contrast media are iodine derivatives, and when injected intravenously, they have various effects on the body. Consequently, an understanding of the various properties is essential to ensure the clinical professional can make informed decisions and take actions to minimize risks to patients.

Due to the variations in the active substances used, it is advised that manufacturer's product information sheets are reviewed prior to administration. Absolute contraindications, as given by the manufacturers, could include hypersensitivity to the active substance, in this case iodine, or any of the excipients of the solution. If a patient has previously sustained a severe contrast reaction, then this may result in that contrast media not using in a future examination or an alternative examination being performed. There are other special precautions necessary with the use of contrast agents that are not absolute contraindications but put the patient at higher risk of adverse reaction. These would include a positive history of multiple well-documented allergies such as asthma and conditions associated with renal impairment [6].

Another consideration is for patients who are receiving metformin therapy. Metformin is a drug used primarily in diabetic patients but often in the treatment of polycystic ovaries. As metformin is solely excreted by the kidneys, any reduction in renal function, for example, following the introduction of a contrast medium, can result in reduced excretion; in turn this can lead to a condition known as lactic acidosis. The procedure used to manage patients on metformin therapy varies from department to department. The Royal College of Radiologists (RCR) has published guidance [6] which suggests that unless a patient's renal function is outside of their normal reference range, metformin therapy need not stop. As a precaution, many X-ray departments have ceased metformin therapy for 48 h post contrast injection.

This is in line with advice contained within the British National Formulary [8]. Some continue with this practice and even ask that a normal renal function test is obtained prior to recommencing metformin treatment.

Contrast administration to pregnant patients should occur only in exceptional circumstances as there is a small risk of thyroid suppression in the fetus [6]. X-ray imaging may also be undesirable due to the radiation burden to the fetus. The risk of proceeding would need to be assessed against the relative benefits of the examination. This decision is generally taken following discussion between the referring clinician and the consultant radiologist. Current RCR guidance for breast-feeding is that no special precautions need to be taken.

As iodinated contrast agents can lead to serious reactions (anaphylactic and anaphylactoid), it is good practice to have emergency drugs and resuscitation equipment easily accessible. Patients should be observed for at least 15 min post injection, as this is when the majority of severe reactions occur. They should also remain in the department for 30 min post procedure to ensure that symptoms do not develop.

Present-day nonionic, low osmolar contrast agents are up to ten times safer than those previously used [6]. As a result, the likelihood of a major life-threatening contrast reaction is very small. Published incidence of severe reactions with nonionic agents is 0.04 % and very serious reactions is 0.004 % [6].

The practitioner performing the examination generally undertakes the delivery of the contrast medium following strict, departmental protocols. The overall responsibility, as with any delegation, lies with the medical practitioner who has prescribed it. Within the UK, the administration of contrast is often conducted by radiographers under a Patient Group Direction (PGD). This is a written instruction which complies with regulation and local quality standards. In the Royal College of Nursing document entitled, "Patient Group Directions – Guidance and Information for Nurses," [7] a PGD is described as an instruction relating to a prescription-only medicine (POM) which provides guidance as to the correct method of administration for an appropriately trained health-care professional (HCP). PGDs are utilized following formal sign-off by the delegating physician and agreed by a pharmacist. They allow designated health-care professionals to administer prescription-only medicines following an assessment of the patient without the requirement of a prescription produced by a medical practitioner or, other independent prescriber.

The management of the risk of complications following the administration of contrast medium is of paramount importance. Prescreening of the patient for the relative contraindications (see below) should reduce the possibility of complications arising. Other good practices include the use of the smallest dose possible to achieve the best quality diagnostic image, the use of premedication and the cessation of nephrotoxic drugs for the 24 h preceding the contrast examination. The use of low osmolar contrast medium in the lowest dose possible while maintaining patient hydration before and after the examination will all assist in the management of the risk of reaction to the contrast medium [9]. Simple steps such as the preheating of contrast medium before administration can assist in the reduction in the number of reactions as this process reduces viscosity [10].

Preexamination checks will highlight the patients at greater risk of allergic reaction to contrast medium and subsequently reduce the risk of an adverse event. All patients who are to receive contrast medium should be checked for:

- Previous contrast medium reactions
- Asthma
- Renal impairment
- Diabetes mellitus
- Metformin therapy

In most cases, the above information can be checked directly with the patient. A positive response to any of the above questions will not necessarily lead to a cancellation of the procedure. Instead, this will allow the prescribing medical practitioner to make an informed decision about the relative risks of administration against the risk of an adverse event. Generally, the final decision to proceed with administration would lie with the radiologist supervising the examination.

The team within radiology should consist of professionals who are skilled in identification of symptoms of contrast medium reaction and trained medical practitioners who are able to competently deal with severe contrast media reactions. This is often the crash team contacted via a local emergency phone number; however, it is essential that the facilities and consumables required for management of adverse reactions are readily available within the department for immediate use.

Any adverse reactions should be accurately recorded within the body of the radiological report to provide a permanent record for future examinations. This will aid future decision making with regard to the techniques used, alternative imaging modalities or premedication. It is worth noting that the value of premedication to minimize the risk of severe contrast medium reaction is questionable [11]. A systematic review completed by Tramer et al. demonstrated a lack of evidence to suggest that routine premedication would assist in the prevention of severe allergic reactions.

Contrast-induced nephropathy (CIN) can also occur as a result of the administration of contrast medium. CIN can be defined as the decline in renal function within 72 h of the administration of contrast medium without any other known cause. This can result in acute renal failure which is clearly very important clinically and can be very costly due to potential for extended length of stay and/or additional requirement for treatment. There are various strategies that can be used in an attempt to manage the risk, and this primarily centers around the identification of the high-risk patients. The incidence of CIN with LOCM in the general population is <2 %; however, in high-risk groups, that is, diabetics or patients with renal impairment, the incidence of CIN is thought to be between 12 and 50 %. Published mortality rates with CIN suggest that if CIN occurs, 16 % of patients will die within 30 days and 25 % within 12 months. If CIN does not occur, the mortality rate is 3.2 % within 30 days and 1.2 % within 12 months. Prescreening of patients undergoing contrast medium injections should occur to minimize the risk. Identification of the high-risk patients and acting accordingly should help to reduce the incidence of CIN. An audit undertaken at a Manchester (UK) teaching hospital identified that of 100 CT

request cards scrutinized, none of the requests included the renal function of the patient. Subsequent analysis showed that 64.5 % of these patients had abnormal renal function (eGFR < 89 ml/min) at the time of imaging. Screening of patients for risk factors is essential; patients with renal impairment, diabetes, hypertension, and gout or those taking nonsteroidal anti-inflammatory (NSAIDs) drugs or diuretics are at higher risk.

12.1.3 The Management of Adverse Reactions

While the steps identified above will minimize the risks of using contrast media, adverse events will inevitably still occur. Subsequently, advice must be readily available for patient management following a reaction.

Possibly the most common reaction is nausea and/or vomiting which is generally managed by providing support, but in prolonged cases of vomiting, antiemetic medication should be considered. Other reactions are rare and include:

- Urticaria
- Bronchospasm
- Laryngeal edema
- Hypotension
- Anaphylactoid reaction

Management of the above ranges from antihistamine administration through intramuscular adrenaline to summoning the crash team.

Psychological preparation of the patient is an essential part of any examination, and patients warrant an explanation of the examination to be performed. As well as improving patient compliance, a thorough explanation of the procedure enables the patient to give the necessary informed consent.

All contrast reactions, regardless of severity, need to be clearly documented so that this information is readily available should the patient attend for a further contrast examination. The information is usually stored on the patient's electronic records/case notes. The patient should also be advised, if they have had a reaction to contrast, that they should make imaging staff aware of this information at any future X-ray appointments which may require the administration of contrast media.

12.1.4 The Role of Contrast Medium in CT Scanning

In 1973, Sir Godfrey Hounsfield described the invention of CT scanning in the *British Journal of Radiology*. The development of this technology over the subsequent 37 years has made computed tomography an essential diagnostic tool in the assessment of a wide range of pathologies and conditions.

Table 12.2 The approximate Hounsfield units for various tissues and substances within the human body

Substance/tissue	HU
Air	−1,000
Fat	−50 to −100
Water	0
CSF	+15
Muscle	+40
Liver	+40 to 60
Contrast	+130
Bone	+400

The physics of CT scanning is complex and as such will not be discussed in depth within this chapter. Instead, the chapter aims to discuss the basics of image formation with subsequent reference to the reasons for contrast medium usage.

Intravenous contrast medium is widely used within CT. Approximately 80 % of CT examinations involve the use of IV contrast media; approximately 65 % of CT examinations use oral contrast medium.

Table 12.2 illustrates the approximate Hounsfield units (HU, CT numbers) for various tissues and substances within the human body as calculated in CT.

HU forms the basis for CT image reconstruction and is based upon the attenuation coefficient of individual 3D tissue voxels, which are converted into 2D pixels and displayed within the resultant image. The similar HU of tissue described in Table 12.2 necessitates the use of contrast medium to increase the contrast between two adjacent structures. As contrast medium is iodine based, it inherently has a high atomic number and as such will attenuate the X-radiation to a greater extent than tissue which does not contain the contrast medium.

12.2 Injector Systems

Previously, when CT scan times could be in the order of 20 min for a routine abdomen and pelvis examination, injection of CT contrast was performed by hand. With the advent of multislice CT which has much faster acquisition times, more accurate timing of the contrast injection is required in order that the contrast is maintained in a bolus ensuring optimum enhancement of the tissues being imaged. It is now routine for intravenous contrast to be administered using a high flow rate (between 2 and 6 ml/s) via a pressure injector. This necessitates a cannula of adequate gauge (preferably 18G but no less than 20G) to be sited, preferably within the antecubital fossa.

As well as the obvious complications with contrast medium, there is also a risk of extravasation of the contrast medium at the injection site. Typically, this occurs when poor injection technique or practice occurs but is also more prevalent in certain groups of patient, for example, pediatrics, elderly patients, and patient with underlying cardiovascular disease. The cannula used should be of adequate gauge for the flow rate intended and ideally inserted immediately prior to use. It should be flushed with saline beforehand to check patency ensuring free flow of contrast

medium will occur. The incidence of extravasation following contrast medium has been demonstrated to be at approximately 0.07 % [12]. Severe reactions are only seen when large amounts of contrast medium are injected. The incidence of this complication has increased following the development of CT pressure injectors. It is good practice to observe the injection site for the duration of the injection for signs of extravasation. The injection can then be stopped immediately following extravasation being identified. However, this is not always possible if the scan commences concurrently which would yield a radiation burden to the staff member involved. To combat this, many manufacturers have developed automatic extravasation detection systems for their pressure injectors which minimize human error and the need for a staff member to be present during the injection or scan phase. The severity of the contrast extravasation is proportional to the amount of contrast that has extravasated. Recent studies have shown that moderate to severe reactions are typically only observed when in excess of 50 ml is extravasated [12].

Signs of extravasation include tightening of the skin, pain, and development of swelling underneath the skin. If any of these are observed, it is recommended that the contrast injection is ceased and the injection site examined closely.

Manual hand injections are still used when imaging the brain with contrast agent as the procedure tends not to be dynamic. Contrast enhancement of the brain is optimal between 1 and 3 min following injection of the contrast medium; however, contrast-enhanced imaging of the brain has been successfully achieved up to 1 h following contrast administration [13]. Normal brain tissue will not enhance with contrast medium, but in the event of disruption to the blood/brain barrier, the abnormal tissue will enhance with associated difference in attenuation value of the tissue which will increase the contrast between this and the normal brain tissue. This allows for better characterization of lesions than compared with CT undertaken without contrast.

Hand-delivery techniques can also be used if the cannula gauge does not allow for pressure injector usage; the injection site is at high risk of extravasation or in other delayed examination – for example, CT urogram – when bolus optimization is of limited importance.

12.2.1 Delivery Techniques

Modern-day multislice CT requires accurate, consistent delivery of contrast medium to ensure the best diagnostic quality can be achieved. Vascular organs throughout the body are enhanced to varying degrees by the contrast medium injected. Scanning should occur in the correct vascular phase to demonstrate the organ or pathology which will answer the clinical question. Possibly the most common example of this is demonstration of the liver using the portal venous phase of circulation. Typically, CT imaging of the liver will occur 60–70 s following the commencement of the contrast medium injection. Scanning is performed in various phases when imaging the liver with CT including the arterial, portal venous, and others and appreciable

differences in attenuation values of the liver tissue can be seen. Often imaging of this type will be completed using 70–90 ml of contrast medium injected at a rate of 3/4 ml/s.

Another good example of a delivery technique in common use involves a dual-phase injection when imaging the tissues of the neck. Due to the complex anatomy, a common protocol for imaging of the soft tissues of the neck when investigating a soft tissue tumor is as follows: A dual-phase injection technique is often utilized. Generally, 90 ml is injected, 50 ml at a rate of 2 ml/s followed by the remaining 40 ml at 1 ml/s. The complete injection will be administered in around 65 s. A delay is then built into the protocol so that scanning commences at 85–90 s. This ensures contrast enhancement within:

- Venous circulation
- Arterial circulation
- Tumor blush

This technique allows better tissue characterization than previous methods which often included arterial circulation only.

The optimization of the bolus of contrast is essential to ensure enhanced diagnostic images. The delivery techniques must attempt to minimize the volume of contrast injected while increasing the time period of enhancement.

The continuous development of MDCT has led to a corresponding decline in the use of diagnostic angiography. CT angiography is now widely accepted as a safe, effective, and sensitive alternative to conventional angiography. Conventional angiography still has a role within diagnostic radiology, but typically this is regarded as an interventional tool. A good example of the application of CT angiography is demonstrated in Fig. 12.3. The images demonstrate a basilar tip aneurysm on CT (Fig. 12.3a) and conventional, digital subtraction angiography (Fig. 12.3b). The CT imaging performed prior to the start of the embolization demonstrates the aneurysm neck and subsequently influences the decision on best course of treatment. Had the neck of the aneurysm been too wide, the patient would not have been fit for endovascular repair, and subsequently the aneurysm would have been clipped with invasive surgery. Figure 12.3c was produced using rotational angiography and offers essential information on planning the aneurysm repair.

Perfusion techniques are proving useful in the evaluation of acute stroke and in brain and liver lesions. The most promising area for development is the evaluation of the ischaemic penumbra which may provide accurate definition of the recoverable brain tissue with thrombolytic treatment. As with angiography, delivery of the contrast medium is required at a flow rate of 4/5 ml/s. A volume of approximately 50 ml will usually suffice. CT images are acquired over a period of approximately 1 min, commencing prior to the contrast delivery and ending following the flow of contrast through arterial, capillary, and venous circulation in the brain. The image dataset produced is subsequently processed to produce a series of maps which can provide visual analysis of the brain. Correlation with the conventional CT imaging allows an accurate diagnosis of acute stroke to be made.

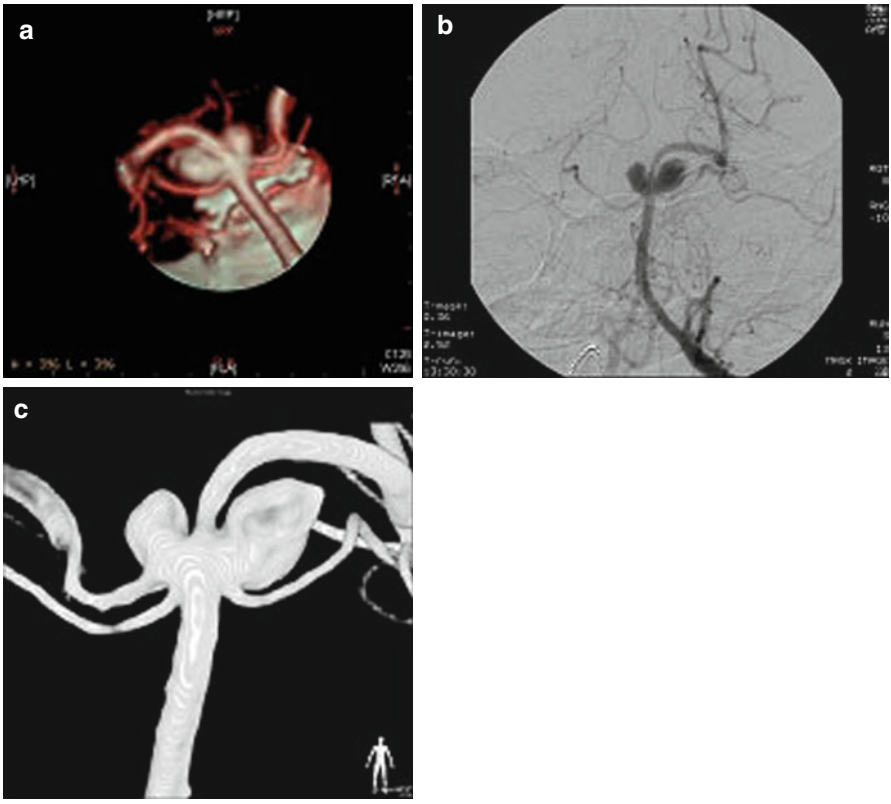

Fig. 12.3 (**a**) CT; (**b**) digital subtraction angiography; (**c**) rotational angiography

Most contrast media can be administered intrathecally; however, in other cases, this practice is contraindicated. This practice is most commonly used in CT myelography, but this technique generally only occurs in specialist neurosurgery centers and in cases where magnetic resonance imaging is contraindicated.

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Chapter 13
Practical Considerations for Performing Clinical SPECT/CT: Nuclear Cardiology; Respiratory; Oncology; Neuroendocrine; Infection; Trauma/Musculoskeletal; Neurological; Miscellaneous Techniques

David Wyn Jones, Marc Griffiths, and Diane Kenwrick

13.1 Introduction

Utilizing a hybrid imaging system for nuclear medicine imaging, as we have seen so far in this book, has many advantages for identifying disease which might be otherwise difficult to localize or characterize. The advantages of SPECT/CT capabilities and values have been discussed at length earlier; however, in this chapter we shall consider the practicalities of using SPECT/CT taking into account practical hints and tips and the possible imaging pitfalls which might be encountered.

With HP-CT systems, the operation of SPECT/CT requires not only the professional expertise of a traditional nuclear medicine technologist but also that of a diagnostic radiographer. Using a HP-CT unit requires additional knowledge to understand the complexity of high-end CT systems, to be able to use the technology safely by optimizing exposures and also to use radiological contrast media to enhance the CT study. Used correctly, CT does not lend itself to being heavily protocol driven as judgments need to be made on an individual patient in order to justify an appropriate exposure which takes into account the presenting medical condition, the physical, and the emotional state of the patient as well as their physical presentation (e.g., size and weight distribution).

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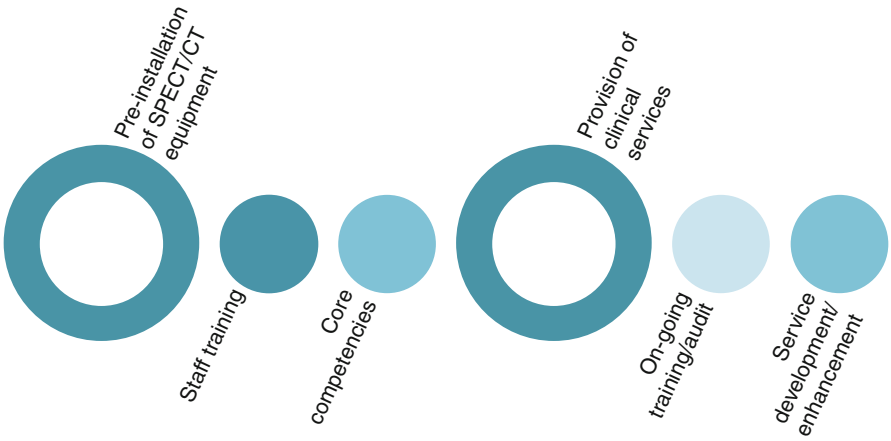


Fig. 13.1 Mapping of staff training and establishing core competencies within SPECT/CT

This chapter will explore some practical considerations that might be of value in the early days of using a hybrid SPECT/CT system. An overview of the scope of practice of professionals working in the specialty is also commented upon, and note is made of where skills and ways of working need to evolve in order to work more effectively with the technology. Common problems and solutions will be presented together with some examples of protocols. This chapter will conclude with some "rule-of-thumb" options in tables to guide practitioners in formulating their own imaging protocols.

13.2 Considerations for Practical Implementation

13.2.1 Appropriate Use of CT in SPECT/CT

The introduction of multimodality imaging within a nuclear medicine department is providing opportunities for the development of new skills and ways of working. The advent of CT hardware and related software has placed greater demands on the abilities of the nuclear medicine workforce and the inclusion of appropriate learning material in the curricula of educational training programmes [1]. A transformational change has occurred within nuclear medicine, in terms of the provision of new technologies informing new techniques and working cultures [2].

Modern SPECT/CT systems can be equipped with multi-slice CT units, capable of providing diagnostic quality anatomical information. During the procurement phase, consideration should be given to the upskilling of the workforce, specifically around the use of CT and the inclusion of relevant audit and competency evaluations post installation. Figure 13.1 gives a diagrammatic representation of the need for staff training and development at the various stages of implementing a new SPECT/CT service.

Fig. 13.2 Color-coded system to identify the possible uses of CT within a SPECT/CT environment

Mapping the initial areas where CT will be utilized with your new SPECT/CT system is essential, and a basic patient workflow mapping exercise could be conducted by members of the departmental staff to achieve this.

Inclusion of new clinical protocols within the daily workflow of the department will help identify "pinch points," especially if there is only one imaging system available, where optimal use of CT can be interspersed with routine dual-phased nuclear medicine imaging and routine gamma camera work. Undertaking an evaluation of the knowledge and skills of the department workforce, along with any projected development needs, is a positive move, demonstrating an investment within the existing staff base. Forging strong links with radiographers who work within dedicated CT departments will certainly help with the in-house training needs of the nuclear medicine workforce. This is especially true in regard of the development of specific problem-solving abilities and the identification of hardware/software problems.

For the majority of nuclear medicine departments adopting "SPECT/CT," the role of fusion imaging is still being scrutinized and implemented according to local imaging requirements [3]. Care should be taken when relying on the standard protocols that are often preset by manufacturers as they may not be optimal for the study or the patient. Assessing patients' previous clinical imaging history prior to a SPECT/CT examination may assist in identifying the appropriate level of CT to use, reflecting the justification of the clinical request. With justification in mind, a recent contrast-enhanced diagnostic CT examination is likely not to warrant another CT scan within the SPECT/CT environment at a high spatial resolution.

Work conducted by Buck et al. [3] provides a systematic approach to the use of low-dose CT (1–4 mSv) within a SPECT/CT environment, which is sufficient for anatomical referencing (localization) and attenuation correction purposes. Specific SPECT/CT protocols should be established for individual patient cases, and these could also include the use of low-dose CT for treatment monitoring. The use of color-coded examinations could also help establish systematic and appropriate workflows within the SPECT/CT environment (Fig. 13.2). Where a "color code" is attributed to the "purpose" of the CT examination, for example, green for stand-alone CT examination or red for attenuation correction only – where the technical factors for the examination can be individually tailored and optimized to the purpose of the particular examination, and the color coding scheme clarifies this to the imaging team.

Clinical departments introducing SPECT/CT services need to ensure that new imaging protocols reflect the varying roles of CT in the imaging procedures locally and that the quality of the CT data is appropriate for the transmission imaging, for example, low-quality data for attenuation correction calculations versus high-quality

Table 13.1 Typical DLP values for CT examinations performed in a 16-slice SPECT/CT system

Clinical area	Typical DLP value for CT examination (mGy · cm)
Sentinel lymph node	20–25 (attenuation correction (A/C) and localization)
Octreotide	80–100 (A/C and localization)
Cardiac	20–40 (A/C)
Skeletal	30–40 (localization)
Brain	255 (A/C)
Diagnostic skeletal, e.g., lumbar spine	300–600 (diagnostic)

imaging required for the fusion of small lesions. Typically, a large percentage of clinical nuclear medicine departments using SPECT/CT will have an established protocol-driven approach with regard to the use of CT using dose modulation; however, the use of this facility needs careful consideration as it can sometimes add to the patient dose. Assistance from medical physics experts will help ensure that radiation doses from CT examinations are optimized and reduced where possible clinically.

Specific dose length product (DPL) values will vary according to the size of individual patients, and different manufacturers will provide details of dose-reduction techniques. In addition, higher specification CT units (e.g., <16-slice systems) will have greater flexibility in terms of being able to provide dedicated CT examinations which may form part of a one-stop service. Cardiac imaging is an example of one-stop services, with the addition of CT calcium scoring to a conventional myocardial perfusion imaging examination.

Table 13.1 provides examples of average DPL values for CT examinations within a SPECT/CT environment.

Additional consideration is required where nuclear medicine departments possess different SPECT/CT systems with different CT specifications. The emergence of a mixed specification SPECT/CT installation base is now evident as a number of institutions look toward upgrading their early generation SPECT/CT systems.

Practitioners need to ensure an appropriate level of understanding with reference to the different capabilities and limitations of the local range of CT units. The majority of CT units installed within integrated SPECT/CT systems offer a single-, dual-, or six-slice acquisition capacity, which can offer similar imaging and acquisition characteristics. However, the introduction of 16-slice CT units (and higher) and flat panel detector units begin to demonstrate different parameters with regard to acquisition modes and patient preparation considerations.

For example, an appreciation of the different appearances of CT image data of the thorax between a single-slice (or LP-CT) and 16-slice CT (HP-CT) hybrid imaging system will be noticeable from the patient's breathing being apparent on the images during the acquisition on one and not the other. Appropriate training and experience is required in order to detect motion artifacts and registration issues, especially where quantification information is presented (e.g., myocardial perfusion imaging).

Incidental findings on CT images need local arrangements for considering the ethical issue of how to report these or highlight the possibility of unexpected findings to the referring clinician and the patient. An example of this might be during a low-resolution

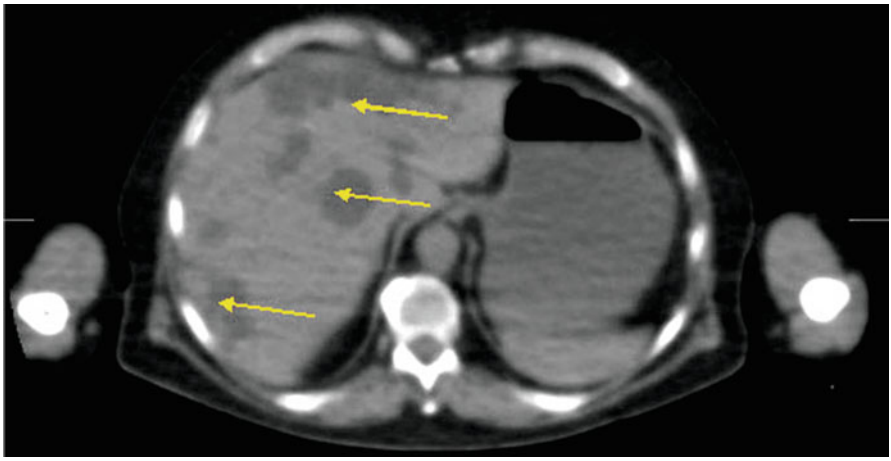


Fig. 13.3 Liver cysts identified on a low-resolution CT exam during a SPECT/CT procedure (*yellow arrows*)

CT scan performed for attenuation correction purposes, where the purpose of the image data is for the calculation of attenuation differences in the patient. Here, the data might be of sufficient quality to delineate a specific disease process which was unexpected and needs further investigation. An example of this would be a lung nodule seen on the lung field incidentally during an attenuation correction acquisition or a liver lesion seen during a CT scan of the lumbar spine for fusion purposes (Fig. 13.3).

In order to be familiar with CT image appearances and abnormal findings, clinical practitioners working within SPECT/CT departments have highlighted the necessity to undertake additional cross-sectional anatomy training, either in-house or via an academic institution [4].

An appreciation of key CT acquisition parameters, including appropriate kVp, mAs, and dose modulation techniques, is essential for certain SPECT/CT examinations. Higher specification CT units will provide the operator with greater flexibility with regard to acquisition parameters, reconstruction filters, and detector array compositions, compared with single- and dual-slice CT units. A clear understanding of the individual strengths and limitations of different specification CT units will help in terms of optimizing overall patient workflow and image quality parameters.

Documenting dose values, normally in terms of the dose length product DLP or CT dose index (CTDI), is a requirement under current legislation relating to the use of ionizing radiation in the UK [5]. This is likely to be the same in other countries. Recording and verifying multiple SPECT/CT examinations within a clinical PACS environment can also prove difficult unless there is a robust system set up to specifically document hybrid imaging studies. Nuclear medicine departments should also consider how archived data is retrieved from a PACS system and what is available for viewing on lower specification web-based PACS browsers throughout hospital environments.

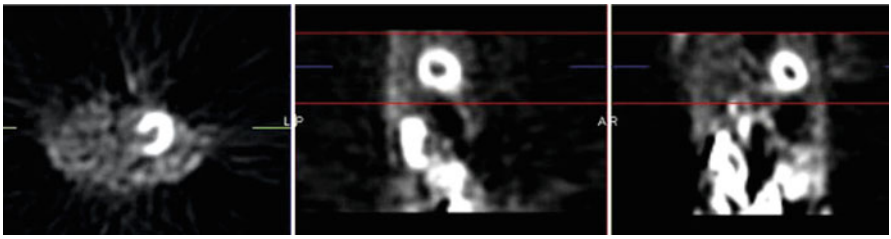


Fig. 13.4 Delineation of CT acquisition limits on guided SPECT myocardial perfusion reconstruction scan

Considerations for lowering patient dose during the CT acquisition of the SPECT/CT examination include:

- Placing the X-ray tube underneath the patient during the topogram. This may reduce the breast dose from 300 μGy to less than 10 μGy [6].
- If available, using an appropriate dose modulation technique during the CT acquisition.
- If available, using the appropriate detector array to acquire the relevant spatial resolution in a minimal time frame.
- Using a "guided SPECT" protocol to acquire the CT acquisition, providing additional confidence in terms of minimizing transmission dose to patients (Fig. 13.4).
- Possible reuse of CT attenuation map for both stress and rest data sets (there are limitations associated with this though).

Gaining appropriate training with regard to the use of a separate acquisition console for the CT examination and unconventional communication methods, such as a voice intercom with patients (Fig. 13.5) and close circuit TV system, should also be factored into the introduction of a SPECT/CT system.

In summary, appropriate use of CT within a SPECT/CT environment has evolved from fixed parameters associated with first-generation systems to a large number of potential imaging parameters. Because of this, systematic training of staff involved in the imaging of patients within SPECT/CT is required, which may consist of in-house training, gaining of experience within a dedicated CT department, and/or the development of clinical competencies, specific to requirements of the nuclear medicine department. There is an opportunity to further develop established clinical roles, with the use of CT at various levels within the SPECT/CT environment. It is vital that appropriate CT training is provided in order to acquire key knowledge and skills for the development of an effective and safe clinical service, and Fig. 13.6 demonstrates some key areas for consideration of training in the various aspects of CT.

Fig. 13.5 Additional operating and voice intercom system utilized for the CT aspect of a SPECT/CT study

13.2.2 Training, Audit, and Standardizing Operating Protocols in SPECT/CT

With the advent of new imaging technologies and techniques, nuclear medicine practitioners are facing the challenge of developing new core skills, a greater degree of multidisciplinary team working, and the emergence of new roles. Hybrid imaging in nuclear medicine has begun to remap the clinical journey for certain patients, and the availability of diagnostic quality CT within the nuclear medicine department has provided the opportunity for service redesign, highlighting the necessity for new knowledge, understanding, and core professional skills. Health-care professionals work within a flux of change and uncertainty, which presents new challenges for the workforce and places an increased demand for new skills and knowledge [7]. Within the UK, the NHS Workforce

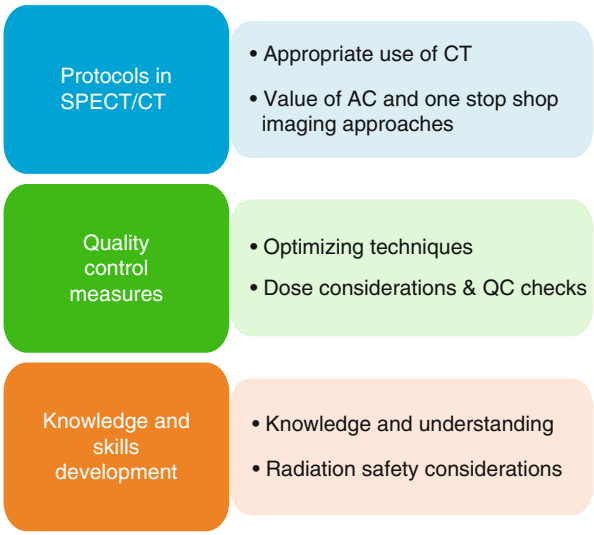


Fig. 13.6 Summary of core training aspects associated with CT within a hybrid imaging environment

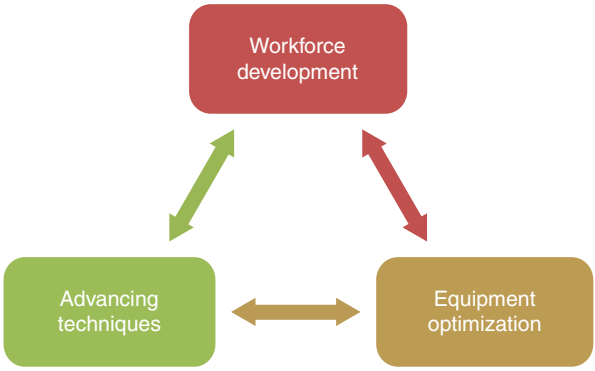


Fig. 13.7 Relationship between balancing the needs of workforce developments and introducing new technology

Review Team [8] has identified the current levels of trained practitioners working within hybrid imaging environments as a national occupational shortage, and clear career trajectories are necessary in order to further develop the workforce.

In the UK, the provision of patient-centered care and core treatment pathways within the modern NHS aims to meet key service improvement strategies to deliver quality patient services [9]. Appropriate training mechanisms are required in order to equip the SPECT/CT workforce, who may need to embed new working practices within the clinical environment. Areas such as cardiology, oncology, and neurology have benefited from the use of CT with modern SPECT/CT systems as a means of demonstrating functional and anatomical within a single environment and, in some instances, a "one-stop" imaging environment for patients. Changes to traditional working practices require careful mapping, evaluation, and quality monitoring process. Optimization of new imaging equipment and emerging techniques requires a skilled workforce, which can adapt and continue to enhance the provision of clinical services. The relationship between introducing the new service and workforce development is shown in Fig. 13.7.

It is important to appreciate the ongoing training needs of the clinical nuclear medicine department and the pace of transformational change that may occur. Despite the technological advancements, it is essential that the patient is at the center of everything and practitioners must continue to use a humanistic approach to their care and treatment. The potential for "objectification" of patients during examinations involving high levels of technology has been reported [10], and it is important that new social frameworks are created in order to permit the acceptance of new working practices [11]. This will enable the creation of new social identities, fostering interprofessional working relationships and further raise the profile of nuclear medicine.

The established knowledge and understanding of first-generation SPECT/CT systems, which includes LP-CT X-ray units, requires a degree of reframing, to take account a range of acquisition and processing parameters. Optimization of SPECT/CT acquisition parameters is essential to current clinical practice, in order to minimize the additional radiation dose from the CT examination and provide an appropriate level of anatomical information, which when justified adds clinical value to the patient's treatment pathway. There is a necessity for clear clinical protocols and appropriate use of CT, especially where the patient may have recently undergone a diagnostic quality (conventional) CT examination.

An increase in SPECT/CT referrals may require revisions to existing clinical protocols and the overall workflow of a nuclear medicine department with some imaging techniques taking longer than before with increasing complexity and the CT component having a stand-alone imaging capability in some variants. This is especially true if there is only one SPECT/CT system available for the departmental workload. Innovative ways of mapping the patient workflow may be necessary, along with the possibility of extending the working day to include more stand-alone CT scanning to justify the additional cost of the unit.

The emergence of SPECT/CT systems being used as backup CT system has also been implemented in some clinical departments. This approach is said to assist the main CT department with their workload and could involve the use of contrast agents, which potentially adds a further layer of complexity within the traditional nuclear medicine department.

During the procurement of a SPECT/CT system, it is vital that staff training, service redesign, and patient workflow dynamics are included within the change process. Staffing a modern SPECT/CT system requires a skilled and competent workforce, who should have the opportunity to further develop their working practice and clinical service provision. Figure 13.8 provides a summary of the main training considerations discussed within this section of the chapter.

13.2.3 Imaging Pitfalls and Considerations for Optimal Clinical Practice

The introduction of an integral CT component has placed considerable demands on the existing design of current gamma camera equipment, with sequential X-ray-based units moving on the same gantry as the gamma camera(s). The provision of a separate

Fig. 13.8 Workforce considerations associated with the introduction of a SPECT/CT system

gantry for the X-ray unit places less mechanical stress on the gamma camera; however, the practitioner needs to be aware of potential imaging pitfalls in order to ensure an optimal service provision.

There are a number of potential imaging artifacts which may emerge within a SPECT/CT imaging environment. Such artifacts can be related to the equipment, patient movement, operator error, and data management issues. The logical steps in producing a diagnostic outcome of sufficient quality are highlighted in Fig. 13.9, which shows the various key stages which may present technical problems within a typical SPECT/CT examination. It is important that the nuclear medicine practitioner possesses an appropriate level of knowledge and understanding, especially when using a transmission-based source to acquire anatomical information. The SPECT/CT workforce should possess context-specific problem identification and solving skills [12].

SPECT/CT data need to be processed, registered, and some form of quality control checks performed before the patient leaves the nuclear medicine department, in order to archive optimal data sets for reporting. The imaging practitioner should also consider using a written method to identify potential imaging artifacts to the person who interprets the images (e.g., metal hip prosthesis). Identification of patient-related artifacts will likely include questioning the patient and/or reviewing their case note. This may assist in minimizing the number of recall examinations within the department; it also provides an opportunity for greater communication with the patient, prior to the examination.

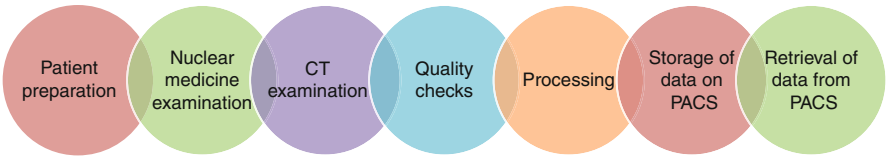


Fig. 13.9 Typical key aspects of a SPECT/CT examination, highlighting the potential for a "cascade of errors" which may amplify as the imaging procedure is performed

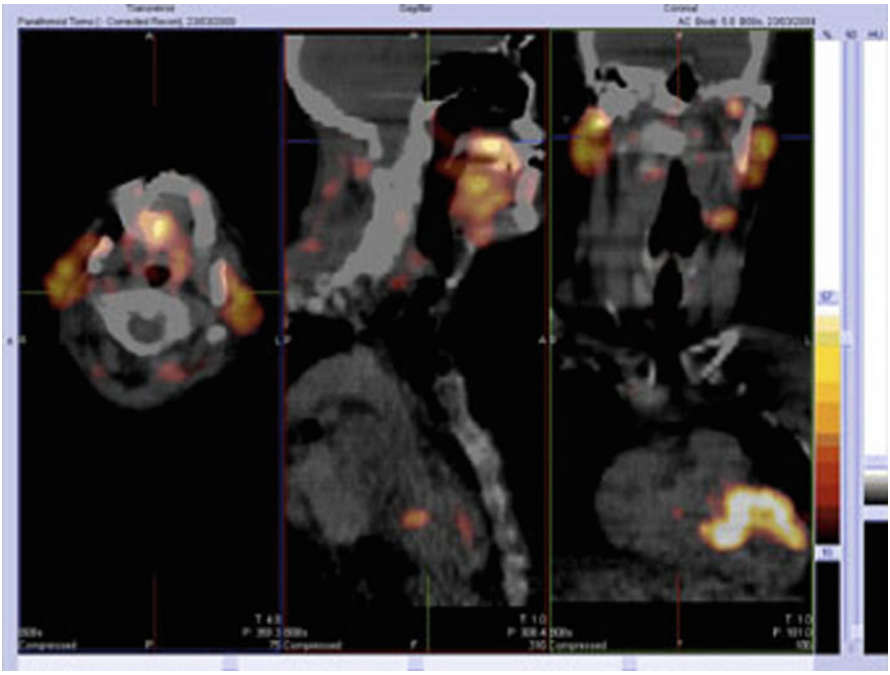


Fig. 13.10 Patient movement during a parathyroid examination, resulting in the misregistration of data

During the SPECT/CT examination, it is crucial that the patient does not move between the emission and transmission acquisitions. One of the most common errors within hybrid imaging practice results from the misregistration of functional and anatomical image sets. Figure 13.10 demonstrates an example of misregistration within a parathyroid examination. It is considered good practice to review the SPECT and CT data, as well as the registered data [13] as artifacts may sometimes be evident on one of the three display options (notably SPECT or CT or registered SPECT/CT).

Modern software systems allow users to perform nonrigid fusion techniques which may correct for the effects of misregistration [13]. Misregistration of SPECT/CT data is more likely to occur at the boundaries of organs, which makes localization difficult to achieve. Published research has demonstrated that even relatively

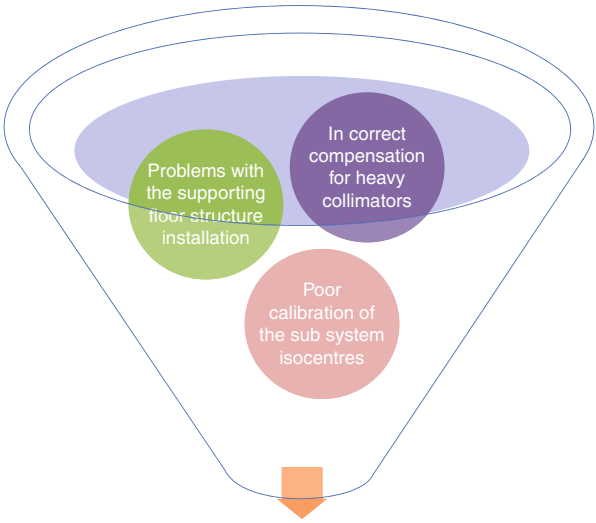


Fig. 13.11 Co-registration issues

small alignment errors (e.g., 1 pixel) in cardiac SPECT/CT can produce a change of 15 % in relative regional activity [14].

Modern SPECT/CT systems have the capability of minimizing the effects of misregistration with accuracy rates of between 0.7 and 1.8 mm [15]. Appropriate quality control checks should be performed to ensure the imaging couch does not "sag" between or during the SPECT and CT aspects of a hybrid examination. Imaging table sag may be due to a number of factors but is predominately related to any of the following in Fig. 13.11.

The use of corrective software may also be employed in cases where gross movement has been identified; however, compensating for the effects of respiratory movement during SPECT/CT can be difficult.

Sequential-based CT units, which were introduced within first-generation SPECT/CT hybrid systems, acquire the transmission-based data over a longer period of time, compared with multi-slice spiral CT systems. CT data could be acquired over a period of four to five respiratory cycles per full rotation with sequential-based hybrid systems, potentially introducing imaging artifacts in thorax studies [16]. Various techniques may be utilized to minimize the effects of respiratory motion during SPECT/CT examinations, including shallow breathing during the CT scan, deep breath hold, and use of a projective-space correction method [16] based on data acquired on a NCAT phantom [17] and processed using an established algorithm [18].

Sufficient access to the patient changing area needs to be considered within SPECT/CT room design specifications. Patients undergoing SPECT/CT procedures should be changed into a standard hospital-based gown and routine questioning undertaken to ensure all removable metallic objects are removed. Failure to remove high-density objects prior to the SPECT/CT examination may result in CT streaking artifacts (see Fig. 13.12) which not only compromises the overall appearance of the images but also

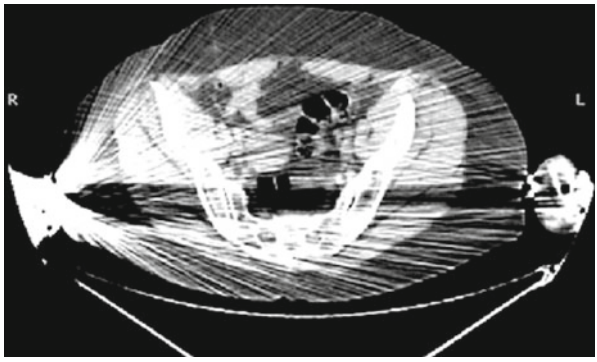


Fig. 13.12 Example of a streaking artifact during a CT examination, caused by metallic bangles and a watch on the patient's wrists

has the possibility to adversely affect the attenuation correction map calculation. Increasing the kVp or producing thinner CT slices may help to compensate the effects of metallic artifacts; however, the potential patient dosimetry implications should also be considered.

Attention should also be taken to ensure the patient is positioned correctly on the imaging couch. Items such as patient gowns and blankets need to be checked and tucked in, as interference with the rotational components of the SPECT/CT system may result in a loss of sensitivity and/or result in them becoming wedged. The practitioner should ensure the imaging couch is free from any objects which may interfere with the mode of travel during the SPECT or CT aspects of the examination. The use of appropriate immobilization devices should be factored into any examination. A full yet suitably worded explanation to the patient is essential because this will assist the patient in complying with the demands of the examination thereby minimizing the occurrence of artifacts on their part.

The addition of CT within the nuclear medicine environment has resulted in the creation of a larger confined space which may present some problems for claustrophobic patients. Patients who are claustrophobic will require extra reassurance before the examination commences; this may help minimize movement artifact. Patients may also be reassured by the use of a webcam or CCTV positioned at the rear end of the SPECT/CT system, which can be used to monitor their progress through the CT component of the hybrid imaging together with the capability of talking to the patient through the intercom system to further reassure during the procedure.

Multi-slice SPECT/CT may also be utilized for stand-alone CT procedures, as a means of providing secondary CT support within the radiology department. Additional work may place greater demands on the CT X-ray tube, and there is a necessity to perform extra quality control tests (e.g., uniformity and image noise), which will need to be factored into the daily and weekly workload. Operators performing CT examinations should possess the relevant skills and knowledge to be able to identify fundamental hardware failures, such as ring artifacts (Fig. 13.13), X-ray photon starvation or truncation artifacts, where part of the patient is outside of the imaging field of view. As previously mentioned, an inconsistent X-ray-based

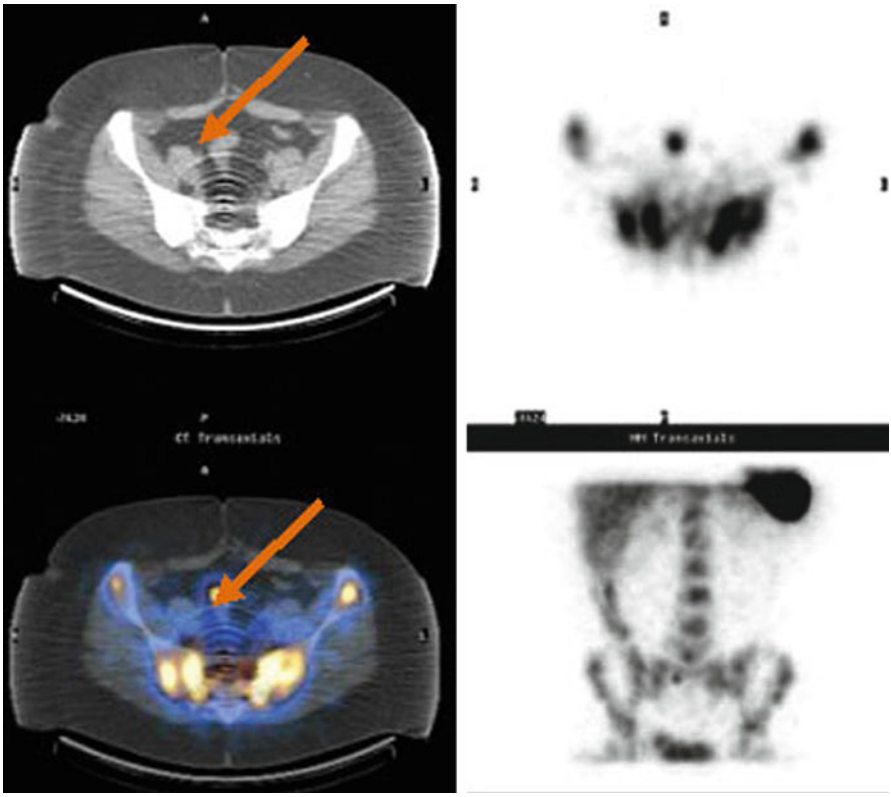


Fig. 13.13 Ring artifact in the central field of view (shown by *arrows*) of the CT scan, which has been subsequently registered on the SPECT/CT scan

transmission photon flux may impact on the attenuation correction map used to correct the SPECT data for the effects of attenuation. This is of particular importance in cardiac-based examinations (potential to mimic appearance of an uptake defect) and dosimetry-based techniques.

There are a number of solutions which may help to minimize some of the aforementioned imaging pitfalls. For example, the use of dose modulation techniques, which automatically varies the X-ray tube current during the course of each rotation minimizes the effects of photon starvation.

Unusual appearance, such as the example presented in Fig. 13.14, may require some critical evaluation of working practice and further communication with the patient and is not to be *confused* with a misregistration artifact.

The use of automated processing workflows provides an environment where patient throughput can be maximized and the management of numerous data sets can be simplified. However, care should be taken to ensure that each of the separate components of the "automated workflow" system and data-processing protocols which have been inserted is scrutinized, to ensure some form of quality control process.

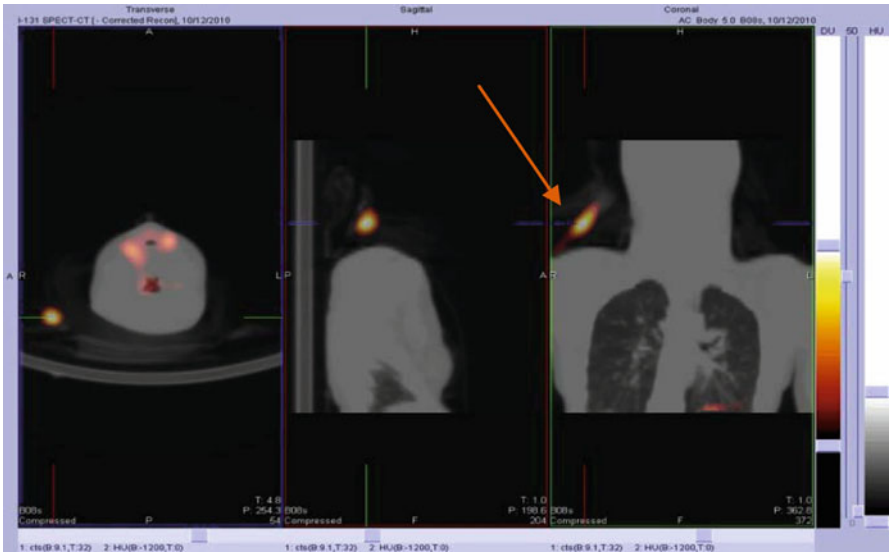


Fig. 13.14 Contamination artifact during an I-131 scan, which was later identified as saliva on the patient's hair (*orange arrow*) not to be confused with a misregistration error

Figure 13.15 provides an example of misregistration in a SPECT/CT examination, which resulted in an incorrect fused data set being sent to PACS.

Correction of misregistered images is possible as a post-processing function with SPECT/CT data with various manufacturers using different methods to achieve a "best-fit" image set or using manual manipulation as appropriate.

Some clinical departments only archive selected processed captures (usually as bitmaps) to their PACS systems, in order to reduce the overall file storage size of the examination; however, sending both SPECT and CT data sets fully might allow "end user" fusion activities and image manipulation and is preferable. Although local copies of every SPECT/CT examination should be kept (e.g., local DVD-R), the additional time required to manually import and reprocess any images could be unproductive should this be required regularly.

13.3 Practical Suggestions for Clinical Use

13.3.1 General Tips on Patient Preparation

When a SPECT/CT study is requested, the referrer must provide adequate clinical information to the practitioner to enable justification of the examination. Although the benefits to the patient outweigh the radiation risk, every effort should still be

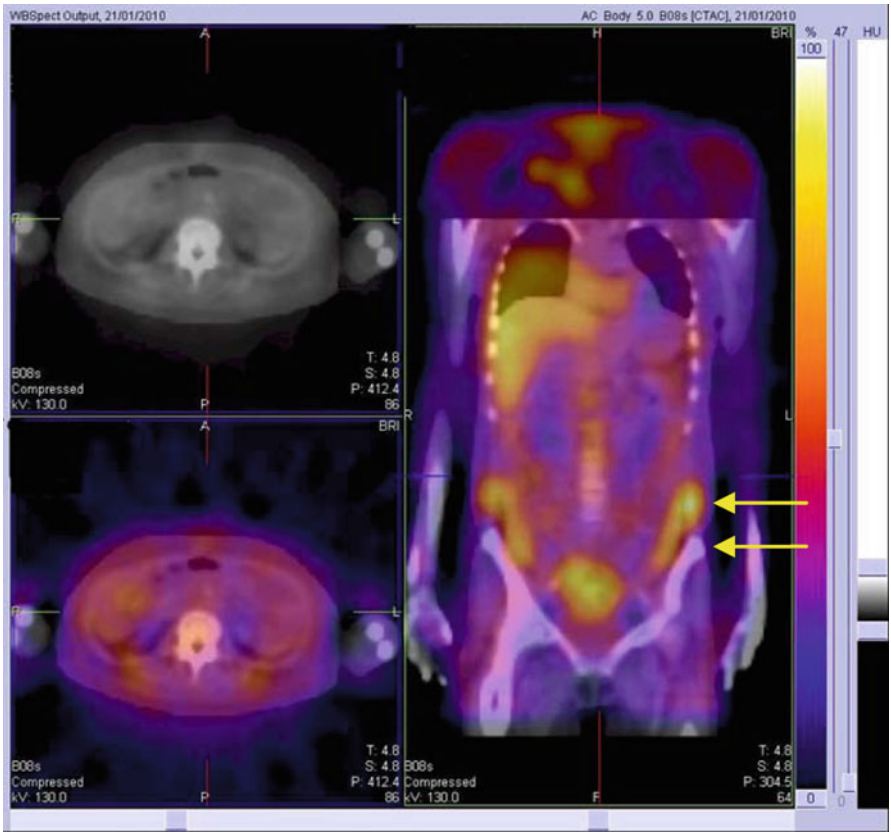


Fig. 13.15 Misregistration of SPECT/CT data sets following incorrect imaging parameters being utilized (*yellow arrows*)

made to minimize the radiation dose [19]. Exposure factors may be adjusted to reduce this dose depending on the indication for undertaking the CT scan [20, 21].

The patient should be identified in accordance with departmental procedure before undertaking the examination. For women of reproductive capacity, the departmental procedure should also be followed; enquiries should be made to establish whether she "is" or "maybe" pregnant or whether she is breast feeding.

As mentioned earlier, prior to undertaking the procedure, it is advisable to ask the patient whether they have any monitors, pumps, prostheses, pacemakers, other surgically implanted devices, body piercing, and, if relevant, any dental work, for example, fillings/metal dentures [22]. Experience shows that careful observational verification and confirmation by double-checking with the patient can avoid metal artifacts ruining the examination. Patient clothing should also be checked; however, it is preferable to standardize this process by asking relevant patients to wear an examination gown thus removing the risk of artifacts from metal items on clothing.

The patient's weight should be checked to ensure it is within the limit for the imaging couch.

As discussed earlier, the procedure can be daunting to claustrophobic patients, so it is important to establish whether a patient suffers with this condition or if they have any other known disabilities or conditions, which could affect the outcome of the study. If the patient is reluctant to undergo the examination, it may be helpful to allow them to lie on the imaging couch and try out the position before administering the radionuclide. It might be the case that the CT part of the examination is the limiting factor (in that the patient will not enter the CT gantry but might tolerate the gamma camera part of the study).

Prior to undertaking the investigation, also ensure that any specific instructions regarding medications, for example, bowel cleansing preparation and oral contrast, have been followed if applicable to the CT part of the investigation. If applicable, check that thyroid blockade has been appropriately administered.

If intravenous contrast is to be used, enquire if the patient has had any recent diagnostic tests with contrast media, for example, barium studies which might preclude the CT taking place without enough time for barium removal – this can take many days. Ask about any known allergies and obtain information about renal function as IV contrast media can cause allergic reactions and deterioration in renal function in vulnerable patients (see Chap. 12).

If carrying out a gated cardiac study, prepare the skin so that there is good electrode contact and ensure that ECG leads will not get caught by any moving equipment.

For dynamic studies venous access should be in situ and accessible to allow administration of isotope and contrast agent without patient movement.

Combining SPECT and CT imaging increases scan acquisition time that potentially adds to patient discomfort and movement. If the patient moves on the bed during and between the scans, spatial resolution is affected and spatial registration of the SPECT and CT images will not be accurately aligned.

Establish whether the patient is able to lie still for the required time which may be between 30 and 45 min. If appropriate, ask the patient if they can raise their arms above the head [20].

The use of positioning aids can be employed to assist with maximum patient comfort and immobilization. Select an ambient room temperature and ensure the patient is warm enough, as a patient can begin to tremble and become agitated and uncomfortable, jeopardizing image fusion alignment, in this scenario. This is particularly important in gamma camera rooms which are well air-conditioned. The thoughtful use of a blanket may seem trivial but can make the difference during a lengthy imaging process.

Patient cooperation is of paramount importance, so ensure the patient understands how the sequential SPECT/CT imaging will occur and emphasize the importance of keeping still during the entire procedure. Explain that if the arms are raised, they must remain in this position throughout.

Inform that patient that to minimize diaphragmatic motion, coughing, sneezing, sleeping, snoring, sighing, yawning, and talking should all be avoided.

Any specific instructions for patient breathing must be explained carefully and may be practiced beforehand to ensure compliance.

If a contour mechanism is operational for the SPECT acquisition, to maximize resolution, there is a need to eliminate any protrusion that may increase detector distance or impede camera rotation, such as the patient's arm position, hair, clothing, or any other items.

Some hybrid systems have the SPECT and the CT detectors mounted on the same rotating platform, in which case the patient should be warned that the gamma camera detectors rotate during CT acquisition.

Artifacts can be due to general patient movement and also breathing; however, they may also be caused by factors not easily controllable by the patient – for example, bowel motility, rapid urinary bladder filling, movement of contrast or radioisotope, prostheses in situ, implants, and dental fillings [22].

13.3.2 Pediatric Imaging with SPECT/CT

Pediatric imaging includes a wide variety of ages ranging from neonates to adolescents. Each age group presents a different challenge, often with no two examinations being the same.

Patients must be correctly identified and for female patients of reproductive capacity follow departmental procedure to make enquiries to establish whether the individual is or maybe pregnant or breast-feeding.

Take a patient history. Ask about medications and allergies and check for any implantable devices and artifacts that may affect the outcome of the procedure [22].

To minimize the radiation dose to the patient, consider the age and weight for both the SPECT and CT scan. If the SPECT/CT system incorporates a multi-slice scanner, try to complete everything required with one examination; otherwise, the CT scan should be restricted to the minimum [20].

Allow adequate time for the investigation and encourage parents and guardians to remain in the room with the patient to help keep them calm, the exception being if the mother/guardian is, or might be, pregnant. Explain that they may not be able to remain in the room for the CT scan unless essential, in which case lead protection will be necessary.

It does help if the department has access to an entertainment system and child-friendly decorations. A favorite toy held or placed just outside of the field of view may provide reassurance, and the use of DVDs can be employed as a distraction to keep a child calm and occupied.

For venopuncture, a local anesthetic applied to the injection site prior to cannulation can make the experience less traumatic and perhaps render the child more cooperative for the image acquisition phase.

Accurate positioning is especially important when imaging small body parts and the patient must remain still [23].

Immobilization aids, such as vacuum pillows, other purpose designed devices, or wrapping with sheets, may be employed, and parents or guardians should be informed of any plan to immobilize [23] and consent verbally as appropriate.

Imaging time may be decreased by reducing the time per step for the SPECT acquisition (use of resolution recovery software may help compensate for loss in image acquisition data), and when the patient's head is not being imaged, they can look to the side so they can see their parents/guardians or a TV screen. However, this may not be possible during the CT acquisition.

To obtain optimum results, sedation may be required and consent will be required from the parents or guardians. Assistance from other health-care professionals will be necessary, and arrangements for sedation should be in place beforehand. If a patient requires pain relief in order to be able to undergo the examination, this should also be arranged in advance [23]. Information may be sent to parents and guardians prior to the test date so they are aware that sedation or immobilization may be required.

Good communication with children is paramount to obtain cooperation. When talking to the child, try to bend down to be at the same level and use understandable explanations. Make sure the child is comfortable and warm throughout the procedure. Younger children and babies may fall asleep on the couch, and dimming the lights can assist with this and also helps sedated children to remain asleep during the procedure.

It is helpful if the child voids their bladder before the examination as this helps to keep them comfortable and still and also takes away an excuse to stop the procedure [23].

For babies and young children, a clean nappy provides comfort and avoids overlying contamination obscuring the SPECT images. Placing absorbent material beneath the patient reduces the risk of contamination to imaging equipment.

On occasions when imaging the pelvis, urinary catheterization may need to be performed by pediatric staff.

13.3.3 Immobilization Aids

Patient movement during the SPECT or CT acquisition affects spatial resolution and causes loss of detail. Patient movement between the SPECT and CT scan affects the spatial registration of the images, resulting in misalignment. Therefore, it is essential that the patient is instructed to remain motionless throughout the procedure without moving between the emission and transmission images.

Consideration toward comfort will assist the patient to maintain position, but additional aids are often employed to ensure proper positioning and immobilization.

Immobilization aids should be radiolucent and suitable for use with gamma emissions and X-rays, antiallergenic, robust, and, for the purpose of infection control, easy to clean or sterilize.

There are many devices available on the market, but most frequently encountered are head restraints (Fig. 13.16), shoulder and arm supports (Fig. 13.17), and leg rests.

Devices that attach directly to the imaging table are less likely to move.

In addition, Velcro strapping, tape, and foam pads can be used (Fig. 13.18).

Velcro straps are often included as part of an immobilization device and may even be incorporated as an integral part of the mattress.

Fig. 13.16 Carbon fiber head rest attached directly onto the imaging table

A simple but effective aid is to place a cylindrical pad under the knees as this often helps to alleviate back ache and reduces the urge to shuffle.

Vacuum pillows or mattresses can also be purchased and inflated around the patient.

Purpose made pediatric immobilization aids are designed to provide safe and comfortable positioning for infants and children, but if not available gentle wrapping in sheets can be used to restrict movement.

It is advisable to tell the parents or guardians of any plans to immobilize beforehand.

To minimize diaphragmatic movement, explain to the patient that coughing, sneezing, sleeping, snoring, and sighing should all be avoided. Talking should also be discouraged, unless essential, and to allow the patient to attract immediate attention, a squeaky rubber toy that can be compressed in the hand may be provided.

13.3.4 Optimal Technique for Correct Fusion Alignment

Even though SPECT/CT systems allow contemporaneous formation of CT and gamma camera images and later fusion, the two data sets are not usually acquired simultaneously [28]. Dependent on design, the patient will need to be

Fig. 13.17 Anatomically designed arm rest

mechanically transported from one imaging component to another using the imaging couch, or even if acquired in the same position on the couch, the component will need to rotate around the subject at differing speeds, therefore requiring a separate acquisition temporally. Also, there might be issues of "cross talk" between gamma and X-ray photons should simultaneous acquisition be possible in the future.

Misregistration of images, as described earlier, can either cause "fusion anomalies," where "active" lesions on the SPECT images are not correctly aligned with their anatomical counterparts on the CT images, in any direction (anterior, posterior, lateral, cranial, caudal) or *suboptimal calculation of attenuation differences* in the

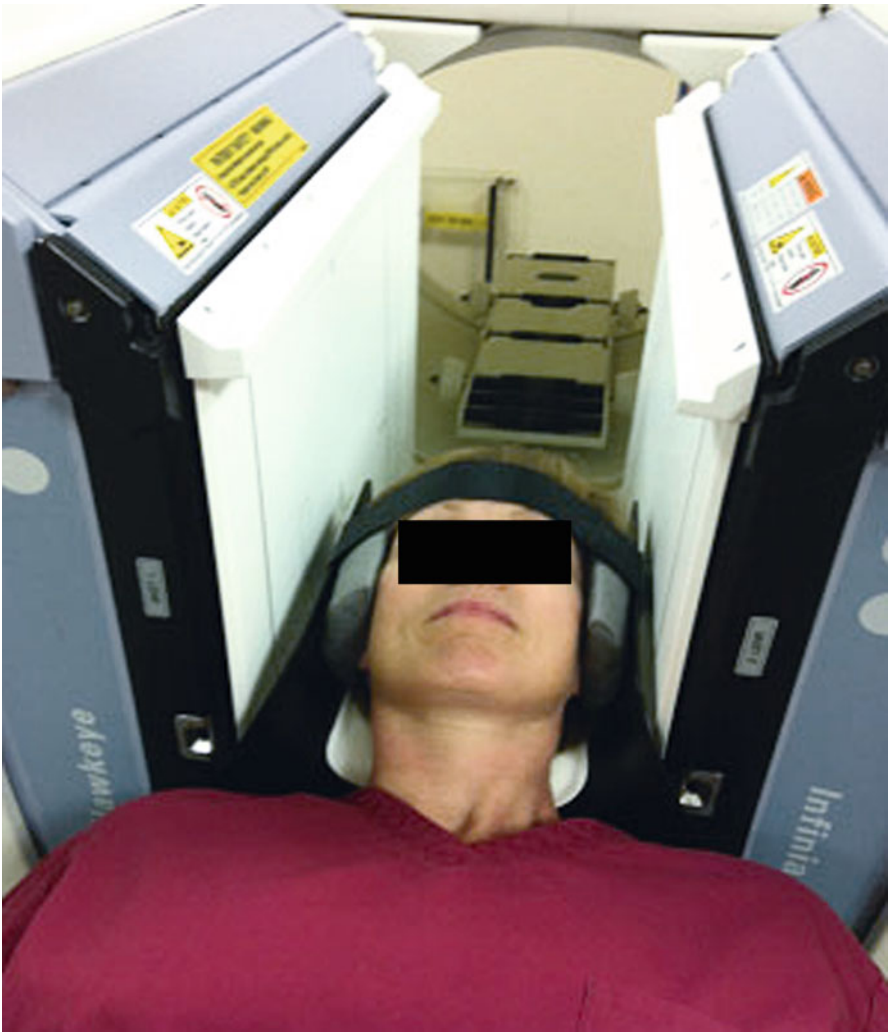


Fig. 13.18 Velcro and foam pads assist with immobilization

body leading to misrepresentation of attenuation losses in the image or in calculated physiological data. Incorrect quantification of cardiac function would be a good example of this.

In order to avoid issues of misregistration, careful consideration of the following aspects will help avoid imaging errors of this nature:

- Respiratory phase
- Patient movement
- Equipment physical features
- Equipment data acquisition factors

In considering whether to ask the patient to stop breathing during CT data acquisition when using HP-CT units (this is not usually an issue for LP-CT units due to the lengthy CT scan), attention needs to be drawn to the nature of the disease being imaged, and it is usually recommended to discuss with the radiologist involved with the case as to whether a breath hold technique is required in order to optimize the quality of the CT images. In this scenario, misregistration of small lesions is possible, as the "spread" of data along the plane of respiration is likely during the long SPECT data acquisition, compared to the relatively short HP-CT exposure in producing fusion CT images. If the patient has already had a recent CT scan, this might not be an issue; however, in the interest of optimizing radiation dose, should a diagnostic quality CT image be required at the time, precise localization with CT of an "active" lesion on the SPECT data might be sacrificed. This is a decision which should be made individually per patient presenting and between imaging team members.

Using a LP-CT unit does not pose this problem (theoretically) as CT exposures are typically longer (over several minutes rather than over several seconds); so there is a natural "averaging" of lesion density over the final image on both components (SPECT and CT) and because the CT data is generally of lower resolution than HP-CT images, conventional CT is still likely to be required for a full diagnostic evaluation of lung parenchyma.

If correct fusion alignment is the primary goal of the examination using HP-CT, then there are some technical factors which can be adjusted to mitigate the apparent disadvantage of the high speed of data acquisition from these units. As well as letting the patient breathe gently during data acquisition consider reducing the gantry rotation speed (e.g., 1.5 s/rotation), so that the overall time taken to cover the scan range is increased or by reducing the number of CT slices per rotation, for example, 16 down to 4 (should the system allow), again increasing the scan duration allowing more averaging of lesion or organ density over the CT image.

Misalignment due to patient movement can be avoided in most circumstances by explaining the procedure fully to the patient and ensuring that they understand the need to remain still for the entire duration of both parts of data acquisition. Some systems require the gamma cameras to rotate during the CT exposure, and this helps the patient to understand that "something is still going on"; however, other systems need to physically move the patient out of the CT component in order to change equipment configuration and then translate the patient back into the system toward the gamma cameras. Experience has shown that patients at this stage can make the wrong assumption that the procedure is over and start to move or make themselves more comfortable, even though the system is only partly through the fusion procedure. Again, reemphasizing to the patient at the transitional stage will help avoid this type of misalignment due to patient movement.

Patients who are in pain will usually find it difficult to remain still enough for a dual acquisition fusion procedure and careful pain management, and the use of positioning accessories, for example, foam pads/Velcro straps, to improve comfort and support will help. Raising the arms to clear the chest is particularly troublesome for some patients due to limited range of movement and the length of the procedure. Consider here reducing the SPECT step times and using

resolution recovery software to regain image resolution with difficult cases. Patients who are too distressed should be rescheduled if possible as although manual alignment is possible should movement occur, this is often impractical should the patient physically contort their body as to twist or bend, this would prove almost impossible to manually realign using manual software adjustments to the final fused image.

One simple patient instruction to void their bladder or use the toilet prior to the procedure is paramount. Not only does this reduce background radiation and remove waste radionuclide from the body but it will also avoid, hopefully, the patient asking to use the toilet between both sets of imaging. Should a patient need to do this – and it is unavoidable – it is possible to salvage the procedure by trying to replicate positioning on the imaging couch and using manual fusion software options – though this is far from ideal. Should the imaging be critical, for example, for radiotherapy planning or possible surgical intervention, consider discussing with the attending radiologist the need to justify repeating either portion of the procedure in order to ensure correct alignment, taking into account the additional radiation burden to the patient. This is usually possible with the long-lived radionuclides, for example, 67Ga/111 In octreotide scans, and procedures where the radionuclide is fixed in location, for example, myocardial perfusion imaging.

Children require additional consideration, and this is discussed in Sect. 13.3.2.

Other factors to consider with fusion alignment are equipment design and robustness for fusion imaging (see Chap. 6) and ensuring that correct fusion alignment is maintained by calibration and quality assurance (see Chap. 6).

When things do not go quite to plan, for example, if the scanner needs rebooting or there is power failure mid procedure, as long as the patient maintains their position on the imaging couch – and they are given plenty of reassurance – then it should theoretically be possible to continue with the fusion procedure, but this is likely to be system specific. It is prudent in this respect to speak to the manufacturer in advance via the support specialist for tips on how to achieve this should the inevitable occur, and a protocol should exist for this locally.

13.3.5 Overcoming Misregistration Issues

Should misalignment be suspected, careful consideration needs to be give to the following points:

1. Prior to declaring the examination complete, the patient should remain still on the imaging couch – and be given a polite reminder – while the images are carefully processed and reconstructed into an initial output suitable for assessing successful image fusion. This should be especially undertaken should patient movement be suspected during data acquisition.
2. Assessing images for misalignment has been briefly discussed in Sect. 13.2.3; however, practical telltale signs are:

(a) Radionuclide "activity overlay" being beyond the borders of the body in any direction when fused on the CT data (Fig. 13.10).
(b) Organs which also show normal distribution of the radionuclide showing misalignment, for example, parathyroid examination using sestamibi showing abnormal distribution in the myocardium as compared to its CT counterpart. The heart should be clearly visible on both data sets to assess alignment; likewise, the liver, and kidneys should be clearly visible on both data sets during an octreotide study to give confidence to general alignment.
(c) Care needs to be taken when declaring a misregistered study, as some of the frames might show some misalignment due to respiration, for example, diaphragmatic motion or peristalsis of stomach and intestines; however, this might be irrelevant to the area under examination, for example, parathyroid glands which might not have been subject to the same movement. In this context movement needs to be assessed in relation to the area under investigation and the likelihood of accessory movement of other organs baring no significance in this regard.

3. Should misalignment be suspected:

(a) Consider whether this is likely to be a technical fault:

(i) Software issue – perhaps reprocess the data sets.

(ii) Check the CT data set – is the "offset" value set at "0"? Some systems allow this to be electronically adjusted pre-exposure and post processing is possible to rectify on some systems.
(iii) Has a manual override function been selected in the viewing package allowing erroneous manual manipulation of the fused data sets?
(iv) Is the system calibration for alignment up to date or needs repeating?

(b) Consider whether this is likely to be a patient movement fault:

(i) If so, can manual manipulation of the data sets faithfully reproduce the pathological condition? Perhaps, view the images and discuss with the reporting clinician – are the manually fused images clinically convincing especially by ensuring that body borders are aligned on the images in three planes?
(ii) If not, is repeating the procedure possible and if so is it justified? Issues such as the radiation dose already given are likely to make a repeat CT scan a fraction of the total dose for the study, and the seriousness of the presenting condition, for example, malignancy or requiring accurate planning for surgery or radiotherapy, might not only render the examination important from a sensitivity point of view but also in relation to the accuracy of delineating focal disease required for intervention.
(iii) Should the procedure be repeated, are you sure that technically the same error will not occur, or will the patient remain still for the extended period of time? Further discussion with the patient as to the likelihood of cooperation will help the technologist assess the situation and decide whether abandonment is the safest scenario.

Table 13.2 1.1 Hounsfield unit, sample CT numbers for various tissues [25]

Tissue	CT number (HU)
Bone	1,000
Liver	40–60
White matter	~20–30
Gray matter	~37–45
Blood	40
Muscle	10–40
Kidney	30
Cerebrospinal fluid	15
Water	0
Fat	−50 to −100
Air	−1,000

13.3.6 The Importance of Correct Windowing

A SPECT/CT examination is carried out for the purpose of combining anatomical and physiological information and provides information that is not usually available from other imaging modalities.

As discussed in earlier chapters, the CT scanner uses a set of software algorithms to determine the amount of X-rays absorbed within every element in a plane of tissue. The tissue attenuation value of each pixel is expressed in Hounsfield units (HU), and a gray scale is then assigned, with the densest structures appearing white and the least dense appearing black [24]. See Table 13.2.

Window width represents the range of the Hounsfield units displayed, and it incorporates levels of a gray scale that is far beyond the ability of the eye to distinguish between. Therefore, just a small portion of the range is displayed. The window level is the number in the center of the window width.

Usually, a system has preset window levels, and widths that can be selected to best demonstrate the anatomy of interest and the gray scale can be adjusted by the user to accentuate anatomic detail and visualize small differences between normal and abnormal tissue.

The nuclear medicine SPECT image display often uses the entire counts per pixel range present in the image [25]. The range of possible pixel values may be represented by a gray scale or a color scale, and most systems incorporate display tables that associate the value of the number of gamma rays to a specific intensity within a color display range.

The image display may also be manipulated by changing to a nonlinear scale [26] in order to limit the range of pixel values displayed and increase the contrast in one region of an image. However, this may be at the expense of other areas within the image where contrast is diminished. This process is known as windowing or thresholding.

Adjustment to contrast enhancement is important to demonstrate normal and abnormal areas within an image when the difference between pixel counts may often be small compared to the maximum counts in the image [27].

Using the available system display levels, pixels containing high counts may be set to the highest level and saturated, while pixels containing low counts may be set to a minimum level and suppressed.

To display the CT and the SPECT images simultaneously, the process of image fusion is applied to achieve registration of a tomographic emission image with a CT transmission image.

Typically, the SPECT data is displayed in color, superimposed onto the CT data that is displayed in the gray scale. The user must manipulate the scales of both imaging modalities to best demonstrate the area of interest and answer any clinical questions. Select an appropriate window for the CT scan depending on whether bone, soft tissue, or lung is to be viewed, and different amounts of SPECT/CT blending may be applied. Inadequate windowing of SPECT, CT, or on the fused images can lead to misinterpretation of the images, as can an unsuitable color selection [20]. An example of this would be imaging of small parathyroid adenomata – where optimal windowing of the color SPECT overlay is of paramount importance to showing very small lesions. Lesions which are small or discrete can be "over-windowed" as to obscure the small lesion or "under-windowed" such as to remove the lesion all together. Check if the patient has had any previous examinations and ensure that a comparable image display is used.

13.3.7 PACS Issues and Practical Solutions

Bringing together two separate imaging components into one specialized medical imaging device poses many challenges, one of which being how to archive and display the output of these complex investigations. The new paradigm in image display introduces new challenges in displaying and manipulating the output of these complex procedures [29]. Workstation interpretation software needs to be capable of full image navigation for the end user.

Not only will conventional nuclear medicine images be produced from SPECT data but also a set of fused images containing the data from the SPECT study merged with data from the CT study undertaken on the same unit. To complicate matters further, consideration needs to be given to archiving and displaying in a useful format, the output of HP-CT data, if the resolution is suitable for conventional diagnostic interpretation. In this respect, the CT data displayed in its conventional format should be output to the picture archiving and communication system (PACS), and if possible this should be represented in the listing as a conventional stand-alone study. For example, should a bone SPECT/CT study of a lumbar spine be undertaken, as well as archiving the SPECT and fusion images separately under the PACS listing for the imaging episode, it is also recommended that the conventional diagnostic quality CT data is output and stored under its own PACS episode listing for a lumbar spine CT scan. This might be problematic, with some current systems not supporting workflows with two PACS procedure identifiers (*accession numbers*); however, it is prudent to try to achieve this by amending the CT accession number if possible

to its own study listing for a CT study and outputting the data to the PACS system after this change. Using this method or similar will prevent end users accessing the PACS system and not realizing that a conventional CT has already been undertaken as part of the fusion study, thus potentially avoiding repeated examinations using ionizing radiation.

Another challenge considering PACS systems is whether it is capable of having full hybrid display functionality at the end user viewing terminal(s). It is desirable to be able to display the fused images together on one visual display unit (VDU), in three orthogonal planes, with cine or scroll functionality to localize lesions simultaneously and be able to fully adjust the gray and color scales to "window" appropriately. Alpha blending (the process of combining two colors to create a third [30]) helps to adjust the "fade in and out" of the fused data sets from one to the other, and this is a crucial function to "reveal" underlying structures on the CT image otherwise overlaid by the SPECT data showing the precise location of the pathological condition. Many current PACS systems do not have these functions available to end users routinely, or they are sometimes costly "add-on" packages available to supplement existing systems. It is prudent to discuss these requirements with information specialists locally prior to procuring future PACS systems.

Practical solutions in the *absence* of full end viewer functionality:

- Simultaneous orthogonal viewing:
  - Output the SPECT data as a volume to the PACS system
  - Output the CT data as a separate study to the PACS system
  - Output a "fused" data set as a volume to the PACS system
- Alpha blending not possible on PACS viewing:
  - The imaging technologist can select appropriate single images (slices) in each plane and save the image as a bitmap (single slice) with and without the SPECT data overlay. This should assist the reviewing clinician by having the precise image with and without SPECT overlay to delineate the pathology precisely; see clinical example in Fig. 13.19a, b.

13.3.8 Optimal CT Data Acquisition and IV Contrast Techniques in SPECT/CT Imaging

As HP-CT hybrid systems become more widespread, it becomes increasingly important for clinical teams to consider radiological imaging and cancer staging in the context of the clinical facility available. Clinical imaging pathways in this regard should consider at which stage a CT scan might be required and if nuclear medicine imaging is likely within the pathway, to consider optimizing radiation dose and resource use by combining the study. This requires sophisticated planning within

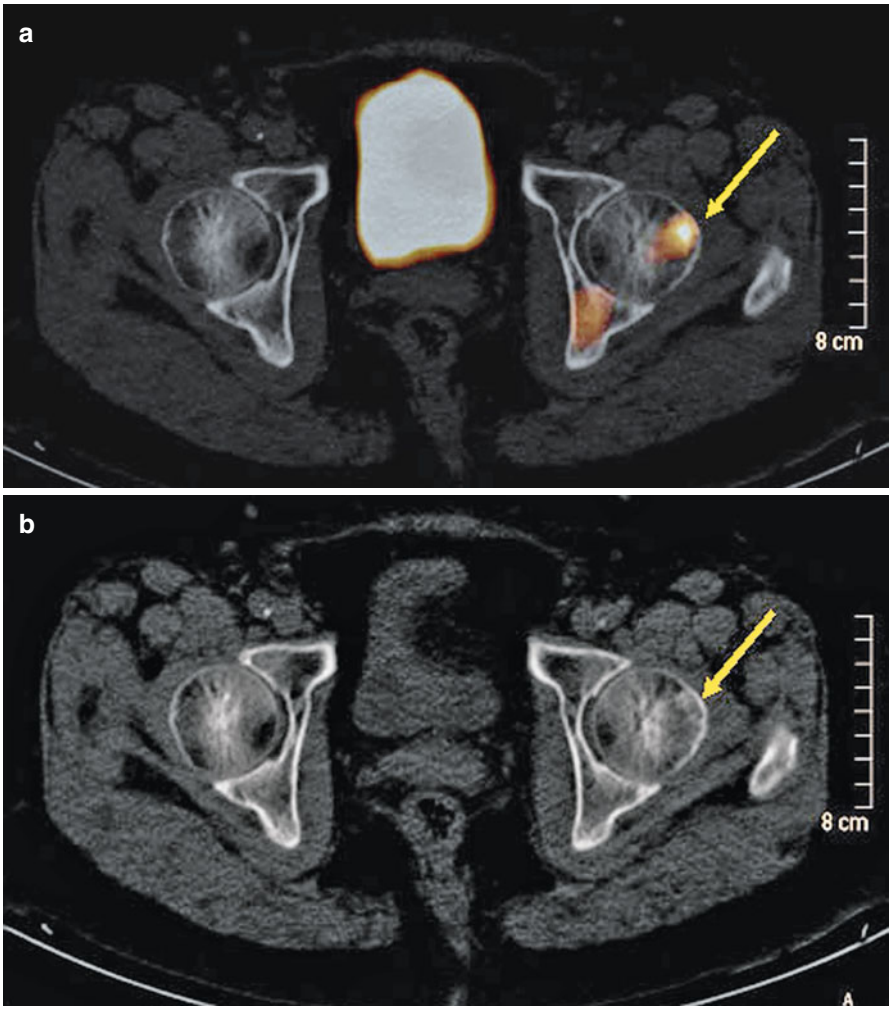


Fig. 13.19 (**a**) *Arrow* shows a fused SPECT/CT bone lesion in left hip. (**b**) *Arrow* shows an area of bone abnormality with SPECT overlay removed

the specialist teams referring patients for imaging. A good example would be neuroendocrine tumor imaging where nuclear imaging is commonplace and often follows a CT scan to initially delineate any obvious pathology. Another example would be HMPAO brain perfusion imaging, where a CT scan of the brain is often undertaken prior to referral.

The above scenarios show how repeat CT exposures can be avoided by well-informed clinical tams, and experience shows that referrals become more sophisticated with increasing local experience of hybrid imaging capabilities.

For this reason, it makes sense to consider HP-CT protocols for fusion in the following situations for optimizing radiation exposures:

(a) Should a relevant CT scan have been recently performed prior to referral for hybrid imaging, then the CT protocol can usually be optimized to produce a volume of data for fusion which is adequate to show location or positional information, where a fully diagnostic quality CT has previously been performed and does not need to be replicated. In this scenario, scan *PITCH*, *slice thickness*, *mAs per slice*, etc., (see Chap. 10) can be optimized to reduce the radiation burden but still provide a valuable data set for fusion. It is prudent to agree this protocol variation in advance between team members and the attending radiologist or physician as a standard technique when CT has recently been performed.

(b) Likewise, should a diagnostic quality CT scan of the patient be planned for the fusion study, then it is usually necessary to produce a CT scan of the same quality as that normally produced with the local conventional CT scanner. In this regard, it is unwise to replicate the settings from another CT scanner as system configuration, and geometry can vary widely, and settings are not readily transferable. However, with careful trial and error, over many initial examinations, the exposure technical factors can usually be manipulated in consultation with the reporting clinician to optimize quality. The use of dose modulation facilities (dose-saving systems) can then maintain image quality (or noise level) with varying patient body habitus (see manufacturer instructions for use on each available system).

Another consideration here is that sometimes the diagnostic CT scan will require intravenous contrast media to enhance tissue contrast and delineate vessels and structures.

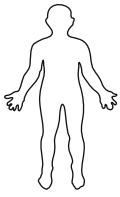
The increasing complexity of hybrid imaging with intravenous contrast agents requires additional knowledge, and this has been the realm of radiology departments mainly until now. However, in order to truly optimize fusion imaging and pathways, it is becoming increasingly necessary for the operators to acquire skills in contrast media use and delivery, with motorized syringe injectors (see Chap. 12). In this way, intravenous contrast media can be delivered, during the CT scan, and then the contrast-enhanced images can be fused with the SPECT data revealing great visual detail and highlighting areas of disease.

The CT data can then be viewed alone as in a conventional CT procedure and also in a fused state. Further advice on manipulating technical factors to optimize radiation dose delivery can be found in Chap. 10.

13.3.9 Example Hybrid CT Technical Parameters

Tables 13.3, 13.4, 13.5, 13.6, and 13.7 give an overview of technical factors which have been shown to be useful at the authors' own clinical center. The factors are system specific and represent what has worked on these machines in the particular clinical setting. NB users are advised to use the tables as a guide only and to consult

Table 13.3 Neuroendocrine abdomen, e.g., Octreoscan/MIBG scan

Examination type/clinical indications: Octreoscan or MIBG scan	Low output CT, e.g., GE Hawkeye	High output CT, e.g., Siemens Symbia/Philips Precedence	Notes Guideline only – consult manufacturer instructions
Tube current mA(s)[a]	2.5	90 mAs/slice	[a]Consider reducing for small adults/children Increase mAs if using narrower slices, e.g., 2 mm
Tube voltage (kV)[b]	140	120	[b]Consider reducing for small adults/children
Slice thickness	10 mm	5 mm	2 mm slices if using data for CT staging also with IV contrast
Pitch	2.6 RPM	Pitch: 0.938 Rot: 0.5 s Collimation: 16 × 1.5	
FOV	40 cm	As selected	Will be less if partial coverage selected
Dose modulation		If possible, e.g., D-DOM	
Reconstruction type, e.g., soft tissue or bone detail		Abdomen/settings C200 W1500	
Expected DLP for average patient	217	App. 300 mGy · cm (routine non-staging)	Will be less if partial coverage is used
Scan area to include	Area of interest	Restrict slices to area of interest (gamma camera coverage) – unless for staging – include whole chest and abdomen	
Technique specifics concerning patient/ positioning/pitfalls	Pt movement/ breathing/ bowel peristalsis/ bladder filling	Allow gentle breathing	If using IV contrast – 100 ml, e.g., Niopam 300 – dynamic for thorax and 65 s post-contrast delay for abdomen images

with equipment manufacturers, application specialists, local medical physics experts, and clinical teams as to the most appropriate settings locally when setting up a new system. Thereafter, it will be the clinical judgment of the operator and/or local physician/radiologist to determine how these factors should be manipulated per patient. Patients come in many shapes and sizes in relation to the examination in question, and people vary in body shape as compared to their BMI. Unfortunately,

Table 13.4 Bone scan fusion

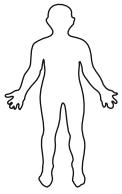
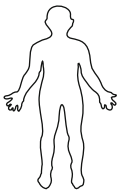
Examination type/ clinical indications: Bone scan	Low output CT, e.g., GE Hawkeye	High output CT, e.g., Siemens Symbia/ Philips Precedence	Notes Guideline only – consult manufacturer instructions
Tube current (mA(s))[a]	2.5	Knee/ankle/foot – 55 Scaphoid 150 Spine pelvis 200 Skull 150	[a]Consider reducing for small adults/children
Tube voltage (kV)[b]	140	Knee/ankle/foot – 120 kV Scaphoid 120 kV Spine pelvis 140 kV Skull 120 kV	[b]Consider reducing for small adults/children
Slice thickness	10 mm	Spiral mode: Knee/ankle/foot – 1 mm slices Scaphoid 0.8 mm Spine pelvis 2 mm Skull 0.8 mm slices	
Pitch	2.6 RPM	Knee/ankle/foot – 0.5 overlap	Generally finer detail, e.g., carpal bones increase overlap (smaller pitch value)
		Scaphoid overlap 0.4 Spine pelvis 1 – no overlap Skull 0.4 overlap	Use maximum possible rotational speed to reduce patient movement
FOV	40 cm		Will be less if partial coverage selected
Dose modulation		If selectable per body area	
Reconstruction type, e.g., soft tissue or bone detail	Bone	High resolution – output bone and soft tissue window settings – fuse on bone settings	
Expected DLP for average patient	217	Range depending on body part, e.g., 70 mGy · cm to 350 mGy · cm DLP	Will be less if partial coverage is used
Scan area to include	Area of interest	Only CT bone area under investigation – unless suspected lesions seen on initial planar gamma camera images – discuss with clinician as appropriate	
Technique specifics concerning patient/ positioning/pitfalls	Rapid bladder filling Pt movement		

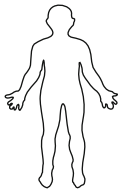
Table 13.5 Myocardial perfusion: CT attenuation correction (CTAC)

Examination type/ clinical indications: MPI	Low output CT, e.g., GE Hawkeye	High output CT, e.g., Siemens Symbia/Philips Precedence	Notes
Tube current mA(s)[a]	2.5	50	[a]Consider reducing for small adults/children – or increasing for very large patient size
Tube voltage (kV)[b]	140	120	[b]Consider reducing for small adults/children
Slice thickness	10 mm	5 mm	
Pitch	2.0 RPM	1[c] (No overlap)	[c]Consider reducing rotational speed of gantry to allow breathing to match SPECT data, e.g., 1.5 s/rotation
FOV	Partial	600 mm (e.g., Precedence)	
Dose modulation		NO[d]	[d]Avoids over penetration of shoulders and increasing dose
Reconstruction type, e.g., soft tissue or bone detail	Mediastinum	As manufacturer recommendation for CTAC	
Expected DLP for average patient	78–116	60–80 mGy · cm	Depends on patient size and heart size
Scan area to include	Myocardium	Myocardium	 Allow increased coverage to compensate for diaphragm motion
Technique specifics concerning patient/ positioning/ pitfalls	Pt movement/ breathing/ cardiomyopathy		

local trial and error will be required to fully optimize CT images according to the level of noise and CT slice geometry that is appropriate to each clinical scenario. Standards have been recommended on the optimal use of SPECT/CT systems by professional bodies [20].

Pediatric CT technical factors are beyond the scope of this text and present a challenge to most imaging departments due to the extra vigilance required in reducing

Table 13.6 Parathyroid glands: needs factors

Examination type/ clinical indications: Parathyroid	Low output CT, e.g., GE Hawkeye	High output CT, e.g., Siemens Symbia/ Philips Precedence	Notes
Tube current[a]	2.5	100	[a]Consider reducing for small adults/children
Tube voltage (kV)[b]	140	120	[b]Consider reducing for small adults/children
Slice thickness	10 mm	2 mm	
Pitch	2.6 RPM	1 (no overlap|)	
FOV	Partial		
Dose modulation		Yes[c]	[c]Optimal data required through hyperdense shoulders
Reconstruction type, e.g., soft tissue or bone detail	Mediastinum	Abdomen/settings C200 W1500	
Expected DLP for average patient	78–102	150–250 mGy · cm	
Scan area to include	Neck and mediastinum Below orbits to mid level of the heart	Neck and mediastinum Below orbits to mid level of the heart	
Technique specifics concerning patient/ positioning/pitfalls	Pt movement/ breathing/ swallowing	Allow gentle breathing – avoid swallowing	Clear throat with water prior to scan

radiation doses to the developing body of the child, allied to the varying body size with age, which is notoriously difficult to judge. In this context, local discussion with medical physics experts and imaging professionals should reach a consensus on how and when to manipulate imaging technical factors and therefore radiation dose.

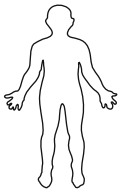
13.4 Conclusion

The user has been presented with a range of information to make informed practical decisions when designing new imaging protocols for undertaking SPECT/CT.

CT does not need to be used in every clinical application, and users would be wise to discuss within their local clinical teams the likely benefit to the patient.

CT used alone to complement conventional nuclear medicine imaging has not been considered in this chapter and is a matter for local discussion. However, many centers now undertake limited CT scanning of "hot" lesions on bone scans in lieu of conventional radiography to characterize pathology.

Table 13.7 BRAIN: HMPAO

Examination type/ clinical indications: Brain	Low output CT, e.g., GE Hawkeye	High output CT, e.g., Siemens Symbia/ Philips Precedence	Notes
Tube current[a]	2.5	280	[a]Consider reducing for small adults/children
Tube voltage (kV)[b]	140	120	[b]Consider reducing for small adults/children
Slice thickness	5 mm	2 mm	
Pitch	1.9/interval 4.42 mm	0.5 (overlap)	
FOV	40 cm	300	
Dose modulation		No	
Reconstruction type: soft tissue or bone detail	Brain/bone	Brain	
Expected DLP for average patient		500–800	Diagnostic quality brain CT – avoid duplication
Scan area to include	From below orbits to skull vault	Below orbits to skull vault	
Technique specifics concerning patient/ positioning/pitfalls			

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Further information on innovation and improvement within the health service is available from:

NHS Institute for Innovation and Improvement: http://www.institute.nhs.uk/.
Quality, Innovation, Productivity and Prevention (QIPP): http://www.institute.nhs.uk/cost_and_quality/qipp/cost_and_quality_homepage.html.
Quality and Service Improvement Tools: http://www.institute.nhs.uk/option,com_quality_and_service_improvement_tools/Itemid,5015.html.
NHS Evidence in Health and Social Care: http://www.evidence.nhs.uk/.

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