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Radiological protection in cone beam computed tomography (CBCT)

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EDITORIAL

ABSTRACT

Radiological Protection
in Cone Beam Computed Tomography (CBCT)

ICRP Publication 1XX

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Abstract-The Commission's *Publications 87* and *102* dealt with patient dose management in computed tomography (CT) and multi-detector CT. The new applications of cone beam CT (CBCT) and the associated radiological protection issues are sufficiently different from those of conventional CT. Thus, the Commission felt it necessary to produce a new document dealing specifically with this technology. The perception that CBCT involves lower doses was only true in initial applications. CBCT is now used widely by specialists who have little or no training in radiological protection. Advice on appropriate utilisation of CBCT needs to be made widely available. Advice on optimisation of protection when using CBCT equipment needs to be strengthened, particularly with respect to the use of newer features of the equipment. Manufacturers should standardise radiation dose displays on CBCT equipment to assist users in optimisation of protection and comparisons of performance. Additional challenges to radiological protection are introduced when CBCT-capable equipment is used for both fluoroscopy and tomography during the same procedure. Mechanisms should be established for tracking and reporting of patient radiation doses from these procedures. Because CBCT technology and applications continue to develop, there are no clear-cut solutions on dosimetry. As a result, the recommendations provided in this publication may evolve in the future as CBCT equipment and applications evolve. As with previous ICRP publications, the Commission hopes that imaging professionals, medical physicists and manufacturers will utilise the guidelines and recommendations provided in this document for the implementation of the Commission's principle of optimisation of protection of patients and medical workers with the objective to keep their exposures low as reasonably achievable, taking into account economic and societal factors, and consistent with achieving the necessary medical outcomes.

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207
208 **PREFACE**
209

210   The International Commission on Radiological Protection (ICRP) provides
211 recommendations and guidance on application of principles of radiological protection it
212 establishes. This has been done through specific publications on the various uses of ionising
213 radiation in medicine in various imaging and therapeutic modalities. This is in addition to the
214 reports published by the ICRP providing advice in general on radiological protection and
215 safety in medicine through Publication 105. Analysis of current technology from point of
216 view of radiological protection has resulted in recommendations directed at manufacturers
217 that have potential for technological developments for safer technology. In this manner the
218 ICRP has acted as an important resource presaging safety issues based on current and future
219 usage of technology and identifying the needs where technology can contribute. Of course
220 there are vast areas of optimization where users can play a big role in minimizing radiation
221 doses to patients without compromising diagnostic or clinical purpose. In recent years there
222 has been evaluation of practice that indicated that large number of imaging procedures have
223 not met the appropriateness guidelines. While ICRP has provided three levels of justification,
224 there is increasing need to scrutinize justification at level 3, and provide guidance on
225 justification of an examination. The current climate of interest in radiological protection has
226 enhanced the audience of ICRP publications to cover policy makers, health authorities, public
227 health organizations, patient groups, organizations developing appropriateness criteria and
228 their use and variety of medical specialists who have now started using imaging technology
229 rather than a decade or so ago. These are all increasing challenges for ICRP publications to
230 address. This publication addresses these challenges in a new technology of cone beam
231 computed tomography (CBCT) that is increasingly being used in day to day practice in
232 hospitals by increasing number of medical specialists. The advise from ICRP is timely. There
233 are issues of patient and worker protection that this publication addresses.
234
235   The Commission launched a Task Group on Radiological Protection in CBCT in 2013.
236
237   The membership of the Task Group was as follows:
238

M.M. Rehani (Chairman)   S. Bartling   R. Gupta

239
240   The corresponding members were:
241

T. Berris (till Oct 2013)   J. M. Boone
G. C. Sharp   R. Pauwels (from Dec 2013)

242
243   Committee 3 critical reviewers were:
244

C. Martin   R. Loose

245
246   Main Commission critical reviewers were:
247

C. Cousins   H.-G. Menzel

248
249   The membership of Committee 3 during the period of preparation of this report was:
250

251   (*2009-2013*)
E. Vañó (Chair)   J.-M. Cosset (Sub-chair)   M.M. Rehani (Secretary)
M.R. Baeza   L.T. Dauer   I. Gusev
J.W. Hopewell   P-L. Khong   P. Ortiz López
S. Mattson   D.L. Miller   K. Åhlström Riklund
H. Ringertz   M. Rosenstein   Y. Yonekura
B. Yue
252
253   (*2013-2017*)
E. Vañó (Chair)   D.L. Miller (Vice-chair)   M.M. Rehani (Secretary)
K. Applegate   M. Bourguignon   L.T. Dauer
S. Demeter   K. Kang   P-L. Khong
R. Loose   P. Ortiz López   C. Martin
K.Å. Riklund   P. Scalliet   Y. Yonekura
B. Yue
254
255

MAIN POINTS

- The guidelines and recommendations on radiological protection in cone beam computed tomography (CBCT) are important because CBCT extends the use of CT to areas that were not typically associated with CT imaging in the past, e.g. surgery, dental and otolaryngology (ear-nose-throat, ENT) clinics, angiography suites, and orthopaedic poly-clinics.

- ICRP's radiological protection principles and recommendations as provided in earlier publications, in particular *Publications 87* (Managing patient dose in computed tomography) and *102* (Managing patient dose in multi-detector computed tomography (MDCT)), apply to these newer applications and should be adhered to.

- Cone-beam nature of the radiation field presents new challenges in dose management to ensure patient safety. The manufacturers of CBCT scanners have invested considerable effort into meeting the electrical and mechanical safety requirements of the users. Similar diligence is needed for issues related to radiation dose and radiological protection.

- This document provides a basis to develop informed decisions and to direct the usage of CBCT for optimising the trade-off between clinical benefit and radiation risk.

- Appropriate use of CBCT, including radiological protection is a joint responsibility of the referring practitioner and the imaging professionals. The imaging professional further has responsibility towards optimisation of protection.

- At the time of writing, tissue reactions from CBCT have not been reported among patients and workers, but growth in usage increases the potential for radiation-induced reactions and injuries.

- Based on recent reports of tissue reactions to radiation, the ICRP emphasises that protection should be optimised not only for whole-body exposures, but also for exposures to specific tissues, especially those of the lens of the eye, the heart, and the cerebrovascular system.

- **The ICRP recommends careful justification for each examination and procedure using CBCT.**

- **The ICRP's concept of "as low as reasonably achievable" should be applied to achieve optimisation within diagnostic reference levels (DRLs).**

- **Since many applications of CBCT involve patient doses similar to MDCT, the room layout and shielding requirements in such cases need to be similar to adequately protect workers.**

- **Traditional CT measurements with a 100-mm chamber are not sufficient for CBCT except for use as internal standard or reference. Dosimetry for CBCT is not yet standardised. Manufacturers should be encouraged to use consistent dose measurement units, and therefore, organisations responsible for establishing radiation units are encouraged to meet the challenge to avoid use of different quantities by manufacturers.**

- **Equipment used for both fluoroscopy and CBCT need to provide aggregate dose indices to individual patients during the entire procedure.**

- **Measurement of dose variables in short phantoms does not provide an accurate indication of the overall dose. But, since determination of the complete rise-to-**

302 **equilibrium dose requires very long phantoms of up to 600 mm, it is impractical to**
303 **perform such measurements in the clinical environment. Therefore, manufacturers**
304 **should measure and provide users with a full set of dosimetric data.**

305 • **Manufacturers should also provide a subset of partial CT dose index (CTDI)**
306 **measurements so that the complete rise-to-equilibrium curve measurements can be**
307 **related to partial measurements that can be performed by users during acceptance**
308 **testing of new equipment. While acceptance tests normally require both phantoms**
309 **and free-in-air measurements, periodic measurement of $CTDI_{air}$ should be sufficient**
310 **as long as free-in-air measurements remain stable with time.**

311 • **Optimisation of both patient and worker doses, particularly when worker has to be**
312 **near the machine, is important wherein monitoring of doses become an essential**
313 **tool. Recording, reporting and tracking of radiation dose for a single patient**
314 **should be made possible.**

315 • **Low dose protocols may be sufficient to answer diagnostic questions focussed on**
316 **high-contrast structures such as lung, bones, dental scans (teeth and maxillofacial),**
317 **ENT scans (paranasal sinuses, skull, temporal bone), interventional material, or**
318 **contrast-enhanced vessels (angiographic interventions).**

319 • **Protocols with higher dose should only be selected if visualisation of soft-tissue**
320 **structures such as intracranial haemorrhage, soft-tissue tumours, or abscesses is**
321 **the primary focus.**

322 • **Most interventional and intra-procedural C-arm CBCT systems can scan an**
323 **angular range spanning 180 to 240 degrees + the cone angle of the x-ray beam. The**
324 **radiosensitive organs, such as thyroid, eyes, female breast and gonads, should be**
325 **on the "detector side" of the arc, whenever possible.**

326 • **Clinical need permitting, every effort should be made by users to ensure the**
327 **volume of interest is fully incorporated in the "field of view" (FOV) provided by**
328 **the CBCT scanners while radiosensitive organs are placed outside the FOV.**

329 • **Post-processing tools such as "thick slice reformats" allow averaging of adjacent**
330 **slices to lower image noise. This may be sufficient for answering certain diagnostic**
331 **questions and evaluation of soft-tissue structures.**

332 • **The aim of CBCT should be to answer a specific diagnostic or intra-operative**
333 **question *vis-à-vis* other imaging modalities and not to obtain image quality that**
334 **rivals MDCT. The decision by the referring practitioner to utilise CBCT should be**
335 **made in consultation with imaging professional.**

336 • **The user must understand the consequences of scan protocol selection not only in**
337 **terms of image quality, but also in terms of applied dose. This is especially**
338 **important for CBCT, where such information may be entirely (and sometimes,**
339 **ambiguously) encoded in the protocol name.**

340 • **There is a need to provide checks and balances, for example, dose check alerts**
341 **implemented in CT in recent years, to avoid high patient doses as compared to**
342 **locally defined reference values.**

343 • **Methods which provide reliable estimates of eye dose under practical situations**
344 **should be established and utilised.**

345 • **The user of CBCT in interventions can significantly influence the radiation dose**
346 **imparted to the patient by judiciously using a "low-image-quality or low dose" vs. a**
347 **"high-image-quality or high dose" scan.**

348 • **In radiotherapy, justified use of CBCT has potential at different stages of therapy**
349 **such as: pre-treatment verification of patient position and target volume localisation;**
350 **and evaluation of non-rigid misalignments, such as flexion of the spine or anatomic**
351 **changes in soft tissue, and during or after treatment to verify that the patient position**
352 **has remained stable throughout the procedure. Low-dose CBCT protocols should be**
353 **used for pre-treatment alignment of bony structures.**

354 • **Many machines were initially only capable of fluoroscopy, but can now additionally**
355 **perform CBCT. Because of the improved clinical information in CBCT, and its**
356 **ability to remove overlying structures, the user may be tempted to over utilise the**
357 **CBCT mode. Users should judiciously use CBCT mode.**

358 • **In orthopaedics, justified use of CBCT can help in assessing the position of fractures**
359 **and implants with respect to the bony anatomy, especially in situations where**
360 **fluoroscopy alone is insufficient and thus help in patient dose management.**

361 • **In urology, low-dose CBCT protocols should be used when imaging high-contrast**
362 **structures, such as calcified kidney stones.**

363 • **Dental CBCT scans should be justified, considering two-dimensional radiography as**
364 **an alternative, and optimised through the use of small FOVs and application- and**
365 **patient-specific exposure factors.**

366 • **The recommendations provided by the Commission on education and training in its**
367 *Publication 113* **are applicable here for CBCT.**

368 • **The level of training in radiological protection should be commensurate with the level**
369 **of expected radiation exposure (ICRP, 2009).**

370 • **All personnel intending to use CBCT for diagnostic purposes should be trained in the**
371 **same manner as for diagnostic CT and for interventional CBCT same as**
372 **interventional procedures using interventional CT.**

373 • **Quality assurance programmes for CBCT should follow guidelines outlined by**
374 **international standards and professional societies.**

375 • **DRLs are not yet established for most CBCT applications. In the absence of**
376 **international or national DRLs, local DRLs should be established to inform local**
377 **policy.**

378

379
380 **GLOSSARY**
381

382 Absorbed dose, D
383   The absorbed dose, D, is the quotient of $d\bar{\varepsilon}$ by dm, where $d\bar{\varepsilon}$ is the mean energy
384   imparted by ionising radiation to matter of mass dm, thus

$$D = \frac{d\bar{\varepsilon}}{dm}$$

385
386   The unit of absorbed dose is J kg^{-1}. The special name for the unit of absorbed dose is
387   gray (Gy); 1 Gy = 1 J kg^{-1}.
388

389 Automatic exposure control (AEC)
390   A device which automatically determines and provides the exposure needed to produce
391   a preselected image quality by sampling the x-ray intensity at the image receptor.
392

393 Collimation
394   Geometrical limitation of the extent of the radiation beam.
395

396 Cone-beam computed tomograpghy (CBCT)
397   A form of x-ray computed tomography (CT) in which the x-rays, in the form of a
398   divergent cone or pyramid, illuminate a two-dimensional (2D) detector array for image
399   capture. Also referred to as digital volume tomography (DVT).
400

401 Dental imaging
402   In this document, dental or oral and maxillofacial imaging refers to imaging of high-
403   contrast structures related to the teeth and jaw bones. Visualisation of other structures
404   (e.g. maxillary sinus, temporomandibular joint, facial skeleton) can be considered as
405   dental imaging if the primary indication for imaging relates to dentistry. Ear-nose-throat
406   (ENT) imaging is considered as a separate application in this document, although it
407   often involves similar radiographic equipment.
408

409 Detector quantum efficiency (DQE)
410   A widely used metric that describes the quality of an x-ray detector. It measures the
411   efficiency (i.e. signal-to-noise performance) of the detector to produce an image from a
412   given incident fluence. Intuitively, it captures how well a detector translates the fluence
413   incident on it into an image, relative to an ideal detector.
414

415 Deterministic effect
416   Injury in populations of cells, characterised by a threshold dose and an increase in the
417   severity of the reaction as the dose is increased further. Also termed tissue reaction. In
418   some cases, deterministic effects are modifiable by post-irradiation procedures
419   including biological response modifiers. Threshold doses for tissue reactions are doses
420   estimated to result in only 1% incidence of tissue reactions.
421

422 Diagnostic reference level (DRL)
423   Dose levels in medical radiodiagnostic practices or, in the case of radiopharmaceuticals,
424   levels of activity, for typical examinations for groups of standard-sized patients or
425   standard phantoms for broadly defined types of equipment. These levels are expected

not to be exceeded for standard procedures when good and normal practice regarding diagnostic and technical performance is applied.

Dose limit
The value of the effective dose or the equivalent dose to individuals from planned exposure situations that shall not be exceeded. Dose limitation is one of three fundamental principles of radiological protection, originally defined by the ICRP.

Effective dose, E
The tissue-weighted sum of the equivalent doses in all specified tissues and organs of the body, given by the expression:

$$E = \sum_{T} w_T H_T$$

where H_T is the equivalent dose in a tissue or organ, T, and w_T is the tissue weighting factor. The SI unit for the effective dose is sievert (Sv), equal to J kg^{-1}.

Equivalent dose, H_T
The dose in a tissue or organ T given by:

$$H_T = \sum_{R} w_R D_{T,R}$$

where $D_{T,R}$ is the mean absorbed dose from radiation R in a tissue or organ T, and w_R is the radiation weighting factor. The unit for the equivalent dose is the same as for effective dose (sievert, Sv), equal to J kg^{-1}.

Hounsfield unit (HU)
Number used to represent the mean x-ray attenuation associated with each elemental area of the CT image. Measured values of attenuation are transformed into HU (also known as CT numbers) using the Hounsfield scale:

$$HU = \frac{\mu_{material} - \mu_{water}}{\mu_{water}} .1000$$

where μ is the effective linear attenuation coefficient of the measured material relative to water for the utilised x-ray beam. The scale is defined so that water has a value of 0 HU and air a value of -1,000 HU.

Justification
One of three fundamental principles of radiological protection, originally defined by the ICRP. The justification principle requires that the net benefit of radiation exposure be positive.

Multi-detector computed tomography (MDCT)
CT scanners with a detector array consisting of more than a single row of detectors. The 'multi-detector-row' configuration of MDCT scanners refers to the use of multiple detector arrays (rows) in the longitudinal direction (that is, along the length of the patient). MDCT scanners utilise third generation CT geometry in which the arc of detectors and the x-ray tube rotate together. All MDCT scanners use a slip-ring gantry, allowing helical acquisition.

Noise

A fundamental statistical phenomenon that is present in all images. Noise tends to reduce the visibility of structures and objects, especially those that have relatively low contrast. In medical imaging, the objective is not to eliminate the noise, but to reduce it to a clinically acceptable level. Noise is the point-to-point variation in image brightness that does not contain useful information. The magnitude of noise is indicated by the standard deviation of the grey values within a region of interest in the image.

Occupational exposure

All exposure incurred by workers in the course of their work, with the exception of (1) excluded exposures and exposures from exempt activities involving radiation or exempt sources; (2) any medical exposure; and (3) the normal local natural background radiation.

Optimisation of protection

One of three fundamental principles of radiological protection, originally defined by the ICRP, defined as: "The likelihood of incurring exposure, the number of people exposed, and the magnitude of their individual doses should all be kept as low as reasonably achievable, taking into account economic and societal factors."

Phantom

A device that absorbs or scatters radiation in an equivalent manner to a patient, utilised to estimate radiation doses and test imaging systems without actually exposing a patient. A phantom may be an anthropomorphic or a physical test object.

Population dose

An expression for the aggregate radiation dose incurred by a population, defined as the product of the number of individuals exposed to a source and their average radiation dose. The collective dose is expressed in man-sievert (man-Sv) and is intended solely as an instrument in the optimisation of radiation protection.

Scatter

Deviation of x-rays from their original trajectory due to interaction with matter.

Shielding

The placement of a high-absorption material (e.g. lead) between the source and its environment, for the purpose of reducing radiation dose to workers, patients or public.

Slice

A tomographic section (defined by the position and thickness) of a test phantom or patient under investigation during a single CT or CBCT exposure.

Stochastic effects of radiation

Malignant disease and heritable effects for which the probability of an effect occurring, but not its severity, is regarded as a function of dose without threshold.

Worker

Any person who is employed, whether full time, part time or temporarily, by an employer, and who has recognised rights and duties in relation to occupational radiological protection.

522
523 **1. INTRODUCTION**
524

525 • The guidelines and recommendations on radiological protection in CBCT are
526 important, because CBCT extends the use of CT to areas that were not typically
527 associated with CT imaging in the past, e.g. surgery, dental and otolaryngology (ENT)
528 clinics, angiography suites, and orthopaedic poly-clinics.
529 • ICRP's radiological protection principles and recommendations as provided in earlier
530 publications, in particular *Publications 87* (Managing patient dose in computed
531 tomography) and *102* (Managing patient dose in multi-detector computed tomography
532 (MDCT)), apply to these newer applications and should be adhered to.
533 • Cone-beam nature of the radiation field presents new challenges in dose management
534 to ensure patient safety. The manufacturers of CBCT scanners have invested
535 considerable effort into meeting the electrical and mechanical safety requirements of
536 the users. Similar diligence is needed for issues related to radiation dose and
537 radiological protection.
538 • This document provides a basis to develop informed decisions and to direct the usage
539 of CBCT for optimising the trade-off between clinical benefit and radiation risk.
540 • Appropriate use of CBCT, including radiological protection is a joint responsibility of
541 the referring practitioner and the imaging professionals. The imaging professional
542 further has responsibility towards optimisation of protection.
543

544 (1)   CBCT is a form of x-ray CT in which the x-rays, in the form of a divergent cone beam,
545 illuminate a wide area-detector for image capture. While conventional MDCT scanners
546 acquire consecutive tomographic slices, in 2D CBCT projection images are acquired by an
547 area detector and directly reconstructed into a three-dimensional (3D) dataset.
548 (2)   CBCT represents an emerging technology that enables high-resolution volumetric
549 scanning of the anatomy under investigation. Just as in MDCT, use of CBCT is steadily
550 increasing in clinical practice. Even though it is a relatively new modality, CBCT is already
551 being used for a variety of clinical applications, such as dental imaging, head and neck
552 imaging (including sinus CT), paediatric imaging, high-resolution bone imaging, and intra-
553 operative and interventional imaging.
554 (3)   CBCT imaging is also used in radiotherapy for pre-treatment verification of patient
555 position and target volume localisation. In this case, the CBCT system is usually mounted on
556 the gantry of a linear accelerator at 90° to the therapeutic beam. For radiotherapy, CBCT
557 imaging is often used for daily repositioning. Under classical fractionation schedules, high
558 cumulative imaging dose to tissues outside the exposure field can accrue.
559 (4)   Although the concept of CBCT has existed for over 25 years, it has only recently
560 become possible to develop clinical CBCT systems that are both sufficiently inexpensive and
561 small enough to be used in operating rooms, out-patient clinics, emergency rooms, and
562 intensive care units. Technological and application-specific factors that have converged to
563 make clinical CBCT possible are:
564   1.   Compact, high-quality flat-panel detector (FPD) arrays;
565   2.   Computer power sufficient for timely cone-beam image reconstruction; and
566   3.   x-ray tubes designed for cone-beam scanning.
567 (5)   Nearly all modern CBCT systems use a digital FPD instead of an image intensifier for
568 image capture. By virtue of these specialised detectors, which are different from the detectors
569 used in conventional MDCT, CBCT is capable of ultra-high spatial resolution and large
570 volume coverage in a single (or partial) rotation of the C-arm. Digital FPDs used in CBCT

571 scanners also enable fluoroscopy, radiography, volumetric CT, and dynamic imaging using a
572 single rotation or partial rotation. These capabilities are extremely useful for intra-operative
573 and vascular applications.

574 (6) The manufacturers of CBCT scanners have invested considerable effort into meeting
575 the electrical and mechanical safety requirements of the users, which are mandated by
576 national regulatory bodies. Similar diligence is needed for issues related to radiation dose. In
577 this respect, the cone-beam nature of the radiation field presents new challenges in dose
578 management to ensure patient safety; guidelines are needed for various stakeholders in this
579 new modality. This report briefly describes the current state-of-the-art CBCT technology,
580 reviews current dose measurement and management approaches, provides recommendations
581 for safe use of CBCT scanners, and identifies gaps that relate to radiological protection where
582 further research is needed.

583 (7) CBCT systems differ from "standard" MDCT systems in several ways that affect
584 image quality and radiological protection. Some key differences are listed below.

585 • Because of the cone-beam nature of the irradiated field and the associated non-
586 uniformities in the primary and scatter radiation imparted to the scan volume, the
587 standard dose metrics popularised by MDCT cannot be applied to CBCT.

588 • CBCT systems have superior spatial resolution for high-contrast objects (e.g. bone,
589 lung) but inferior contrast resolution for low-contrast objects (e.g. soft tissue). A
590 trained and skilled user of CBCT can significantly influence the radiation dose
591 imparted to the patient by judiciously deciding whether a "high-dose" scan is needed
592 or a "low-dose" one will suffice. A high-dose scan is generally required if soft-tissue
593 structures are the main diagnostic focus, while for angiographic scans with arterial or
594 venous contrast media, or for defining the position of interventional catheters, a low-
595 dose scan may be sufficient.

596 • Because of the higher spatial resolution of a FPD, CBCT slices are intrinsically
597 thinner and have lower signal-to-noise ratios (SNRs) for the same dose than MDCT
598 slices. Any attempt to match the SNR in a thin CBCT slice with a thick MDCT slice
599 will result in a proportionate increase in dose. Instead, increasing the slice thickness,
600 or other similar image processing methods, should be applied to improve the SNR in
601 CBCT.

602 • In many CBCT scanners, the angular span over which the projection data are acquired
603 can be customised. This feature is not generally available in MDCT, but can be used
604 in CBCT to minimise the dose to selected organs.

605 (8) The purpose of this report is to identify radiological protection issues for patients and
606 workers and, in line with other ICRP publications, recommendations are set out for all
607 stakeholders ranging from day-to-day clinical users, auxiliary support workers, buyers,
608 manufacturers, and policy directing committees.

609 (9) The primary target audience of this document, as most other documents produced by
610 the Commission related to protection in medicine, is health professionals working with CBCT,
611 or other workers tasked with radiation protection and image quality optimisation in CBCT,
612 manufacturers of imaging equipment, regulators, and policy makers in charge of radiological
613 protection.

614
615 **1.1. History of development**
616

617 (10) The first CBCT scanner was built for angiography at the Mayo Clinic, Rochester, NY,
618 in 1982 (Robb, 1982). Multiple teams in the early 1990s further pursued the idea of multi-
619 angle projections from a wide-area detector for medical imaging. For example, Saint-Felix et

620 al. (1994) clinically tested a system called the Morphometer consisting of two imaging chains,
621 each with an x-ray tube and an image intensifier (Saint-Félix et al., 1994). This CBCT system
622 was designed for 3D angiography using the gantry of a conventional CT scanner. It
623 reconstructed vascular images from a set of digitally subtracted angiography (DSA) images.
624 This gantry platform, which was never released clinically, was abandoned in favour of a C-
625 arm supporting a single imaging chain.

626 (11) Fahrig et al. (1997, 1998) also developed a CBCT system based on an image
627 intensifier and C-arm for use in angiography. Wiesent et al. (2000) developed a similar
628 system comprising a C-arm plus an image intensifier for interventional angiography. Ning et
629 al. (2000a,b) and Wang (1997) developed a CBCT angiography imager based on a GE 8800
630 CT scanner with an image intensifier – charge-coupled device (CCD) chain and later with a
631 FPD. Schueler et al. (1997) and Kawata et al. (1996) developed a CBCT angiography scanner
632 based on a biplanar C-arm system.

633 (12) Jaffray and Siewerdsen (1999, 2000, 2001) developed a CBCT system for
634 radiotherapy guidance based on an amorphous silicon FPD. Efforts are also underway to
635 build a dedicated CBCT-based imaging system for mammography (O'Connell et al., 2010;
636 Packard et al., 2012; Kalender et al., 2012).

1.2. Current standards in radiological protection in CBCT

640 (13) The guidelines and recommendations on radiological protection in CBCT are
641 especially important, because CBCT extends the use of CT to areas that were not typically
642 associated with CT imaging in the past, e.g. surgery, dental and otolaryngology (ENT) clinics,
643 angiography suites, and orthopaedic poly-clinics. Fundamentally, CBCT is a form of CT, and
644 as such, most facility design and quality assurance (QA) requirements that apply to MDCT
645 should also be applied to CBCT. This, however, can lead to an erroneous impression that
646 CBCT is identical to MDCT, making it difficult to manage CBCT from operational and
647 radiation safety points of view. Further complications arise when a user is tempted to regard
648 CBCT as a "light" or "low-dose" CT, a view that is maintained because CBCT functionality
649 is often an adjunct to existing capabilities, such as fluoroscopy and angiography in a C-arm or
650 other clinic-based systems. Embedded in these user biases is the risk for potential overuse of
651 CBCT resulting in unnecessary radiation dose to the patients and/or workers.

652 (14) Traditionally, the use of CBCT in dentistry has entailed a relatively low radiation dose.
653 However, this is not always the case, and many recent applications of CBCT, especially in
654 ENT and interventional procedures, can impart much higher radiation doses that equal or
655 exceed those from MDCT (Dijkstra et al., 2011; Kyriakou et al., 2008; Schulz et al., 2012).
656 There are also situations in which multiple CBCT procedures have to be performed on one
657 patient (such as CBCT-guided interventions) enhancing the need to keep the inflicted
658 radiation dose to a minimum. Therefore, dose implications of CBCT pose a risk from the
659 perspective of an individual patient as well as for the risk from radiation exposure of the
660 population as a whole.

661 (15) Imaging professionals and medical physicists are well aware of the radiation dose
662 issues in CT. This knowledge, however, does not directly translate to CBCT, for which the
663 trade-off between image quality and radiation dose can be quite complex. At the same time,
664 clinical users as well as those undertaking QA and members of radiation safety committees
665 need clear guidelines on operating and regulating these systems. This document, which is
666 presumably the first on radiological protection in CBCT from an international source,
667 provides a basis for developing informed clinical decisions on the usage of CBCT and
668 guidance for optimising the trade-off between clinical benefit and radiation risk.

669   (16) It is worth clarifying the terminology used in CBCT literature, as some of the terms
670   may be used ambiguously. The term "cone beam" in its most basic meaning refers to a system
671   with an x-ray beam that extends "significantly" in the z direction, in addition to the x-y- or
672   axial plane. It is difficult to define how much z coverage is mandatory for a CT system to be
673   called CBCT. At a rudimentary level, all MDCT systems with 16 or 64 rows of detectors are
674   *cone-beamed* CT scanners as they provide 2 to 4 cm of z-coverage. However, the techniques
675   involved in choosing exposure parameters relate to conventional CT scanning and most
676   imaging professionals, engineers and equipment vendors would not regard these MDCT
677   scanners as cone-beam scanners. For the purposes of this document, we will call a CT scanner
678   CBCT if: (1) it is based on a wide-area detector (typically, a digital FPD); (2) has a field of
679   view (FOV) that extends more the 8 cm in the z-direction. This second criterion is empiric
680   and derived from most commonly available platforms at the present time, excluding dental
681   CBCT for which smaller FOVs can be used; and (3) uses a reconstruction algorithm that
682   accounts for the cone-beam nature of the x-ray illumination without resorting to a parallel
683   beam approximation. The last point, by itself, is not sufficient as many MDCT reconstruction
684   algorithms take into account the cone-beam nature of the source along the z-direction.
685
686   **1.3. Responsibilities of different stakeholders**
687
688   (17) Approximately 80 million CT scans are performed every year in the US, and this
689   number is increasing on a yearly basis (Sierzenski et al., 2014). Multiple recent papers have
690   drawn attention to the population dose from these scans (Brenner 2010). There is also
691   increasing realisation that a large fraction of this radiation dose to the population is avoidable
692   as it comes from unjustified or inappropriate examinations. Currently, data on inappropriate
693   use are mostly available for CT rather than CBCT. Appropriate use of CT scanning is a joint
694   responsibility of the referring practitioner and the imaging professional, and most national
695   regulations assign this responsibility either jointly or to the imaging professional. Since a
696   referring practitioner best understands the clinical need for the examination, he/she must
697   interact with an imaging specialist to arrive at the radiological examination or procedure that
698   is in the best interest of the patient. Electronic referrals with decision support have the
699   potential to simplify and streamline this interaction while making this process more evidence
700   based (Sistrom et al., 2009). Such systems can go a long way towards facilitating the desired
701   radiological examination performed with the lowest radiation dose while maintaining the
702   image quality needed for the clinical purpose. Practitioners, technologists and medical
703   physicists must understand their role and responsibilities in this endeavour. To this end, there
704   is need to further develop methods that facilitate the interaction between referring practitioner
705   and imaging professional to translate their joint responsibility for radiological safety into
706   practice.
707   (18) Over the years, manufacturers have played a vital role in technological developments
708   to reduce patient doses for particular CT examinations. The Commission, while
709   acknowledging this role, hopes that manufacturers will remain on the forefront of developing
710   new technologies for radiological protection of patients and workers.
711
712   **1.4. Why is it important to know CBCT doses?**
713
714   (19) It is easy for a practitioner, not versed in the details of dose management, to dismiss
715   CBCT as upgraded fluoroscopy coupled with 3D reconstruction. For the most part, the dose
716   from CBCT is indeed lower than that from MDCT, which may reinforce this belief. However,
717   uncritical application of CBCT under the assumption that it is a modality with minimal dose

718 consequences could result in significant doses in some circumstances and is not appropriate
719 for the protection of the patient.
720 (20) CBCT is a relatively new development in clinical practice. Data on radiation doses
721 and possible effects of CBCT are still being gathered and analysed. Even at this early stage,
722 however, studies indicate that there is room for optimisation to keep the radiation dose as low
723 as reasonably achievable. This report systematically summarises the available dose data
724 related to CBCT use and discusses radiological protection issues for patients and workers.
725 Given the potential of CBCT to become a significant source of radiation dose to patients in
726 the future, it is appropriate to be mindful of the radiation exposure while utilising the full
727 diagnostic potential of this exciting modality. In 1999-2000, while preparing its *Publication
728 87* (ICRP, 2000; Rehani and Berry, 2000), the Commission had similarly presaged the need to
729 watch for increasing radiation doses from MDCT. Although this concern was not well
730 appreciated at that time, it has become a major issue in subsequent years with multiple high
731 profile reports in the media. This publication provides a similar review of the current CBCT
732 literature and presents the data regarding radiation dose to patients and worker in use of
733 CBCT.

735 **1.5. Safety in perspective**

737 (21) Safety is achieved most readily when it is built into the system rather than a matter of
738 choice for users. A good example is a collision avoidance system, an innovation that started
739 in the automobile industry but has been implemented in multiple types of imaging gantries to
740 avoid accidents. With such a system in place, if the gantry of the imaging device comes into
741 contact with a person or object, it simply stops moving. In the absence of such a system,
742 when collision avoidance has to be accomplished primarily via user education, training and
743 instructions, the risk of injury from collisions will be higher. There are instances when both
744 detection of an anomalous condition and its automatic avoidance cannot be simultaneously
745 implemented. In such cases, detection and warning may accomplish a similar end result. For
746 example, radars for detection of speed limits have been shown to decrease the incidence of
747 speeding violations.
748 (22) For radiation safety in MDCT, a display of radiation exposure information on the
749 operator console has often been present for a number of years. After a series of accidental
750 exposures was reported in the US in 2007–2008, MDCT systems can now automatically
751 detect settings to prevent accidental exposure (NEMA, 2010). Such systems provide an
752 additional layer of non-intrusive checks and balances in the conduct of a scan. Display of
753 such information on CBCT consoles needs to be standardised. The Commission recommends
754 development and implementation of safety systems that require the least amount of
755 interaction from the operator and workers while providing:
756 • Regular and continuous monitoring of radiation output throughout the examination;
757 • Automatic comparison with reference or desired dose levels which need to be
758 established;
759 • Timely feedback to the system operator;
760 • Wide availability of automatic adjustment of the dose to a prescribed level in a manner
761 that is somewhat similar to AEC; and
762 • Alerts when dose is higher than specified. Currently, dose check does not apply to CBCT
763 systems (NEMA, 2010).
764 (23) Other technologies that many CBCT vendors need to uniformly implement include
765 automatic collimation control so that the x-ray beam always falls on the detector; guidance

766 for instruments during image-guided interventions, and minimisation of scatter dose resulting
767 from mechanical components.
768
769 <center>**1.6. Scope of the document**</center>
770
771 (24) Since a substantial amount of information is currently available on dental CBCT
772 including a document issued by the European Commission project SEDENTEXCT (Safety
773 and Efficacy of a New and Emerging Dental X-ray Modality) (http://www.sedentexct.eu/), it
774 was decided to restrict the current document to non-dental applications of CBCT, with a brief
775 coverage of dental CBCT.
776 (25) It should be emphasised that the main focus of this report is on doses to patients and
777 workers coming from CBCT acquisitions. CBCT acquisition can be part of fluoroscopically
778 guided procedures. In such cases, the dose from fluoroscopy and relevant implications need to
779 be accounted for. ICRP *Publication 117* included information pertinent to radiation protection
780 of patients and workers in fluoroscopic procedures performed outside imaging departments
781 (ICRP, 2010), and ICRP *Publication 120* covered radiation protection of patients and workers
782 during interventional fluoroscopy (ICRP, 2013).
783
784 <center>**1.7. References**</center>
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855

856
857   **2. CBCT TECHNOLOGY**
858
859   **2.1. Introduction**
860
861   (26) In the past decade, development of digital FPDs for conventional x-ray radiography,
862   fluoroscopy and mammography has propelled the use of CBCT into the mainstream of
863   medical imaging. Most CBCT systems currently in use leverage the power of dynamic FPDs
864   (i.e. able to acquire several frames per second (FPS), as opposed to static FPDs) to provide
865   volumetric 3D datasets.
866   (27) A C-arm gantry consisting of a digital FPD and a large cone-angle x-ray tube is the
867   most commonly used platform for CBCT. There are a number of other implementations of
868   CBCT that differ in the mechanical gantry used for scanning, the detector subsystem, the type
869   of x-ray tube and filtration, the cone angle employed for imaging, and the algorithm used for
870   reconstructions. The following section describes and introduces different types of CBCT
871   scanners.
872
873   **2.2. Technological issues**
874
875   (28) As far as tomographic capabilities of a CBCT scanner are concerned, in simple terms,
876   one can think of them as a conventional MDCT in which the rows of detector elements
877   (typically 16 to 64 rows) have been replaced by an area detector (Popescu et al., 2005; Ross et
878   al., 2004; Grasruck et al., 2005). In general, a CBCT scanner consists of an x-ray source, a
879   detector, and a gantry to move this imaging chain around the patient. We briefly describe the
880   most commonly used subsystems.
881
882   **2.2.1. X-ray source**
883
884   (29) The x-ray source used in a CBCT scanner must provide a broad, cone-shaped beam of
885   radiation. Consequently, CBCT scanners use a much larger anode angle than a tube used in an
886   MDCT scanner. Typical operating conditions are an x-ray tube voltage of 50-140 kVp, a tube
887   current of 10-800 mA, and a total power of 10-80 kW. In order to take advantage of the small
888   detector pixel size, the focal spot size ranges from 0.2 mm to 0.8 mm. The typical FOV
889   covered in one rotation, using a single FPD, can be as much as 25 cm in the angular direction,
890   and 20 cm in the z-direction. Larger sizes are possible when multiple panels or dual scans are
891   used, such that the principle axis of the x-ray illumination is offset from the centre of the
892   panel to allow beam correction.
893
894   **2.2.2. Detector**
895
896   (30) While some older systems still use an image intensifier, most modern CBCT scanners
897   use a digital FPD. FPDs provide higher dose efficiency and dynamic range than the other
898   detector technologies they replaced (x-ray film, film/screen combinations, and image
899   intensifiers); however, their dynamic range is lower than that of standard MDCT detectors
900   (Miracle and Mukerji, 2009). FPDs also generally provide higher spatial resolution than
901   image intensifiers and conventional detector arrays used in MDCT. Direct digital readout up
902   to 30 FPS ensures that the data are available in a directly usable form for both projection and
903   3D reconstruction.

904   (31) The native resolution of a flat panel is typically at or below 200 μm, although higher
905 resolution detector panels are available. After accounting for magnification and x-ray focal
906 spot size, this yields an isotropic voxel resolution of approximately 150 μm. Generally, in 3D
907 acquisition mode, the FPD is operated in a 2x2 binning mode (summing signals from two
908 rows and two columns to increase the SNR and the readout speed, and to reduce the matrix
909 size), and the isotropic resolution is of the order of 200 μm. Therefore, compared to
910 conventional MDCT scanners, a flat panel-based CBCT system improves the spatial
911 resolution by a factor of almost 12 on a voxel-by-voxel basis. Its high spatial resolution is
912 capable of visualising complex human anatomy, including fine structures of the maxillofacial
913 region and skull base.
914   (32) Typically, the FPD used in CBCT is composed of a matrix of detector elements that
915 can span anywhere from 5x5 cm^2 to 40x40 cm^2. Such scanners, therefore, are capable of
916 producing a large number of slices spanning anywhere from 5 to 20 cm in one rotation. The
917 z-coverage afforded by these scanners can be large enough to image an entire organ such as
918 the brain, heart, liver, or kidneys in one axial scan.
919
920 **2.2.3. Gantry**
921
922   (33) Depending on the mechanical system of the gantry, CBCT scanners can allow
923 conventional fluoroscopy, angiography and radiography in the same setup as well as
924 providing high spatial resolution and large volume coverage. These facilities make such
925 machines especially attractive for intra-operative and vascular applications. The various
926 gantry platforms that are commonly used are described.
927
928 *C-arm based CBCT*
929   (34) All major imaging equipment vendors now provide C-arm scanners that employ
930 digital FPDs integrated with a C-arm gantry (See Fig. 2.1.). The C-arm platform offers open
931 architecture and ready patient access. There are two major C-arm based setups that need to be
932 distinguished.
933
934   (35) ***C-arm based interventional CBCT.*** One can use the C-arm for fluoroscopy and
935 projective angiography (including DSA). However, by putting the C-arm in a fast-spin mode
936 while acquiring images, one can obtain projection data that can be converted into relatively
937 high quality, high contrast CT images. Interventional procedures are usually performed using
938 fluoroscopy. The operator can intermittently use the CBCT mode for clarification and 3D
939 localisation (Orth et al., 2008; Schafer et al., 2011). These machines, therefore, enable a
940 seamless integration of these heretofore separate modalities. They are used in angiographic,
941 surgical, orthopaedic, urologic and other interventional settings.
942
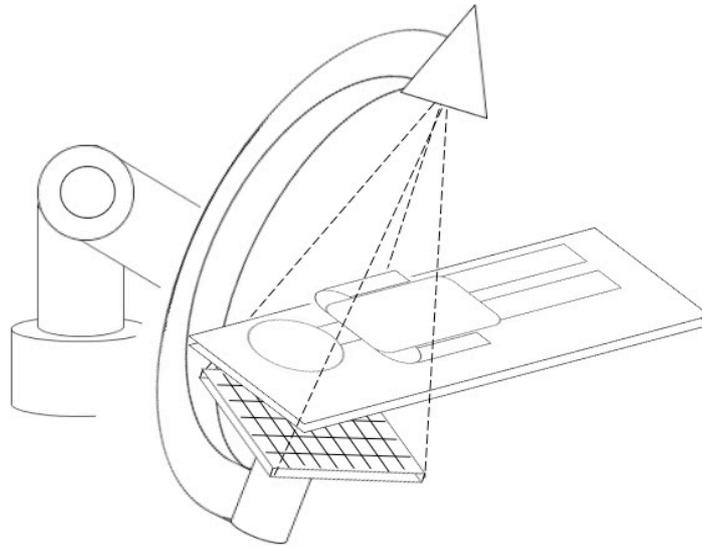
943
944 Fig. 2.1. C-arm based CBCT. A C-arm is used to mount the imaging chain and this provides
945 the necessary amount of freedom required to revolve around the patient. C-arm systems are
946 used in surgical, orthopaedic, urologic or interventional environments (image provided by
947 Rolf Kueres). (permissions required)
948
949 (36) ***Dedicated C-arm based CBCT systems.*** A number of systems dedicated for dental,
950 ENT, head and neck, extremity imaging, and mammography are available. One popular
951 variation of C-arm based CBCT systems is the so-called "seat-scanners", in which a small C-
952 arm, with a horizontal imaging chain consisting of a FPD and an x-ray tube, revolves around
953 the head of the patient while they sit on a chair (Fig. 2.2.). Alternatively, for certain models,
954 the patient is in a supine or standing position. These scanners are dedicated to dental,
955 maxillofacial and temporal bone applications because of their relatively small scan FOV.
956 Besides weight and mechanical considerations, there is no fundamental reason why their FOV
957 cannot be increased. They are currently limited to these niche applications.
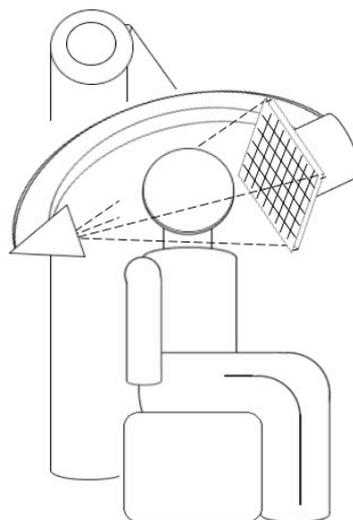
958

959 Fig. 2.2. Clinic-based CBCT systems. The imaging chain is mounted on a horizontal rotating
960 C-arm. These systems are usually used in head and neck applications (image provided by
961 Rolf Kueres).
962
963 *Gantry-based CBCT*
964 (37) A flat-panel volume CT (VCT) scanner combines the advances in CT with digital FPD
965 technology (see Fig. 2.3.). It is in fact a CT machine in which the detector rows have been
966 replaced by a FPD. From an operational point of view, the main difference between a CT-
967 gantry based and a C-arm based cone-beam system lies in basic engineering: the gantry-based
968 systems are more stable and have fewer geometric inaccuracies compared to the C-arm based
969 systems. In addition, the isocentre of any CT gantry, by virtue of its mechanical design, is
970 much more precisely defined than the best C-arm gantries. As a result, gantry-based designs
971 may in most cases offer better spatial resolution.
972 (38) In a C-arm system, the detector and the x-ray tube are connected to the control
973 hardware by an umbilical cord of cables that prevents them from continuously spinning
974 around the patient. This is not the case for a CT gantry-based system, in which a slip ring is
975 used to take data from a rotating component. Elaborate collision avoidance schemes have
976 been implemented to ensure operator safety. No such concerns exist for CT gantry-based
977 systems.
978 (39) By virtue of a FPD, CT gantry-based CBCT systems are capable of ultra-high spatial
979 resolution, direct volumetric imaging, and continuous rotation around a patient. Continuous
980 rotation enables dynamic CT scanning, the ability to observe a process evolving with time
981 such as perfusion of an entire organ such as the brain, liver, or kidney (e.g. after transplant or
982 an ischemic event).
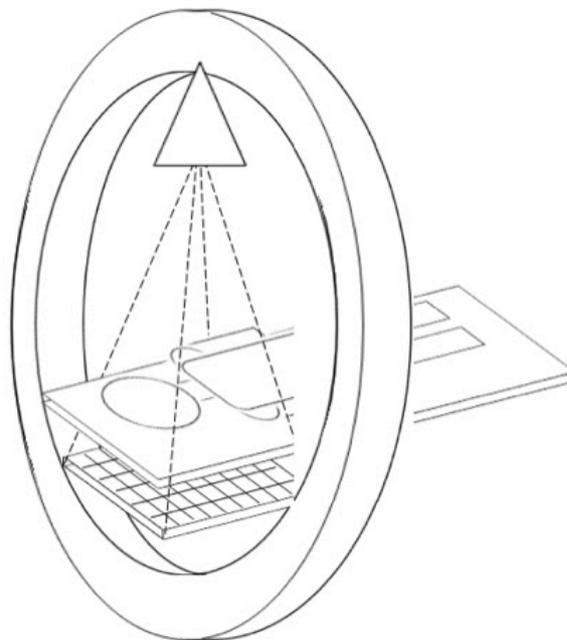
983
984 Fig. 2.3. Gantry-based CBCT. The patient lies on a patient bed, and the imaging chain
985 revolves around the patient like in MDCT (image provided by Rolf Kueres). (permissions
986 required)
987

988 *CBCT in radiotherapy*
989   (40) In radiotherapy, CBCT is used for precise alignment of the target volume with a
990 therapeutic, hard x-ray beam from a linear accelerator. Two separate arrangements, dubbed
991 kV CBCT and MV CBCT, are popular. In kV CBCT, a separate imaging chain consisting of
992 an x-ray tube operated in the kV range is used as the x-ray source, and a FPD is used for
993 imaging. The entire imaging chain is mounted on the linac gantry, in an orientation that is
994 orthogonal to the therapeutic beam. A routine CBCT scan is conducted prior to the therapy
995 for precise alignment.
996   (41) The MV CBCT uses the high energy x-rays from the linac itself for imaging. AFPD
997 that can operate at very high x-ray photon energies is used to acquire the projection data, and
998 a separate imaging chain is not required. Given the high photon energy and associated
999 decrease in photoelectric absorption, the soft-tissue contrast of MV CBCT is markedly worse
1000 than that of kV CBCT. However, it is sufficient to visualise bony anatomy, which may be
1001 acceptable for alignment purposes.
1002
1003 *Co-integrated systems*
1004   (42) Co-integrated systems exist mainly in nuclear medicine (e.g. single photon emission
1005 tomography: SPECT) (Sowards-Emmerd et al., 2009). Here, a flat-panel CBCT system is
1006 mounted on the same gantry as the nuclear imaging chain. The CBCT data are used for
1007 attenuation correction and anatomic localisation.
1008

1009   **2.3. Clinical scenarios where CBCT is used**

1010
1011   (43) In current clinical practice, CBCT scanners are being used for a variety of imaging
1012 applications ranging from preclinical to clinical imaging (Table 2.1.). Their use is primarily
1013 motivated by taking advantage of the following 3 special characteristics: (1) combining
1014 dynamic fluoroscopy/angiography and tomographic imaging; (2) large z-coverage; and (3)
1015 high-resolution imaging of high-contrast structures.
1016

1017 Table 2.1. CBCT in a variety of medical applications ranging from research to clinical imaging.

Application	Setup	Synonyms	Leading advantage why CBCT is used	Use cases	Common use examples of CBCT
Non-vascular interventional procedures	C-arm system	3D C-arm, CBCT	1, 2	Liver intervention, abscess drainage, skeletal interventions	Spatial position control of intervention instruments and material
Vascular head/body interventions	C-arm system	Angiographic CT, Rotational angiography-CT	1	Tumour embolisation, bleeding, revascularisation in peripheral occlusive disease	Spatial position of intervention instrument, rule out of bleeding, embolisation therapy control
Vascular cardiac interventions	C-arm system	Rotational angiography-CT	1	Electrophysiological catheter ablation	Spatial assessment of instrument position
Orthopaedic	Mobile C-		1, 2	Osteosynthesis	Spatial position

interventions	arm/O-Arm systems				of implants, complex fractures
Radiation therapy planning/guidance	Gantry or C-arm (with treatment system)		2	Tumour therapy	Patient registration, physiological motion control
Dental, ENT	Over-the-head C-arm "seat-scanner"/gantry based	DVT	3	Dental workup, paranasal sinus, temporal bone	Diagnostic imaging, datasets for navigation (implantology)
Breast	Horizontal gantry based		2, 3	Rule out carcinoma, biopsy	
Urology	C-arm		2, 3	Lithotripsy, diagnostic workup	Diagnostic imaging, stone detection
Nuclear medicine Hybrid imaging (SPECT/CT)	Transmission and emission systems mounted on rotating gantry		2	Attenuation correction, anatomic localisation (fused physiological and anatomic data sets)	Myocardial perfusion imaging, skeletal imaging, oncology imaging
Peripheral bone imaging	C-arm/gantry based		3	Osteoporosis	Bone microstructures, bone density
Animal imaging/Specimen imaging	Bench-top, gantry based		2,3	Research and veterinary	Experimental imaging

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1042

1043
1044   **3. THE BIOLOGICAL EFFECTS OF RADIATION**
1045

1046 • **At the time of writing, tissue reactions from CBCT have not been reported among**
1047   **patients and workers, but growth in usage increases the potential for radiation-**
1048   **induced reactions and injuries.**

1049 • **Based on recent reports of tissue reactions to radiation, the ICRP emphasises that**
1050   **protection should be optimised not only for whole-body exposures, but also for**
1051   **exposures to specific tissues, especially the lens of the eye, the heart, and the**
1052   **cerebrovascular system.**

1053
1054   **3.1. Introduction**
1055

1056 (44) Effects of ionising radiation are classified into two main categories, based on the
1057 underlying biological mechanism: those that are a result of cell death are called tissue
1058 reactions or deterministic effects. Such effects include skin erythema, hair loss, cataracts,
1059 infertility, vascular disease, and hematopoietic and gastroenterological effects. Those within
1060 the second category, which are a result of cell mutations, are known as stochastic effects and
1061 include cancer and genetic effects.
1062 (45) Tissue reactions appear when the radiation dose exceeds a specific threshold. The
1063 severity of reaction depends on the total radiation dose received by the organ or part of organ.
1064 On the other hand, stochastic effects are governed more by the inherent randomness in
1065 microscopic interactions between radiation and biological matter. In most cancer models, the
1066 probability of cancer induction due to exposure to radiation is considered to be proportional
1067 to the radiation dose. Moreover, for the purpose of radiation protection, no matter how low
1068 the radiation dose, theoretically there is always a small probability that it will induce cancer
1069 or heritable effects.
1070
1071   **3.2. Tissue reactions**
1072

1073 (46) For tissue reactions, the damage to cells is related directly to radiation dose and a dose
1074 threshold exists. ICRP *Publication 103* (ICRP, 2007b) states that; "The reason for the
1075 presence of this threshold dose is that radiation damage (serious malfunction or death) of a
1076 critical population of cells in a given tissue needs to be sustained before injury is expressed in
1077 a clinically relevant form. Above the threshold dose the severity of the injury, including
1078 impairment of the capacity for tissue recovery, increases with dose". Tissue reactions have
1079 thresholds that are typically of the order of few hundreds of mGy. Skin effects may occur at
1080 absorbed doses of 3 Gy; threshold doses for other organs are provided in Table 3.1.
1081 (47) As a classical example, erythematous effects commonly occurred on the workers'
1082 hands during the early days of radiology, about a century ago. Such symptoms have rarely
1083 happened in the last 50 years in workers using medical x-rays. However, skin injuries have
1084 been observed among patients due to fluoroscopic procedures in interventional radiology and
1085 cardiology (ICRP, 2001; Balter et al., 2010; Rehani and Srimahachota 2011; ICRP, 2013).
1086 Also, in interventional procedures, problems including hair loss and chronic occupational
1087 dermatitis have been reported for radiologists and cardiologists on body parts unprotected by
1088 the lead apron or lead table shield (Wiper et al., 2005; Rehani and Ortiz López, 2006). To the
1089 best of our knowledge, there have been no reports to date of skin injuries in patients
1090 undergoing CBCT. Regarding MDCT, skin injuries have been observed in the past few years

1091 in patients undergoing MDCT scans, mainly as a result of inappropriate use of scanners
1092 (ICRP, 2007a). Hair loss has been reported among patients undergoing brain perfusion CT
1093 (Bogdanich, 2009; Bogdanich, 2010; Wintermark and Lev, 2010). Although skin injuries
1094 related to CBCT have not been reported among patients or workers, the technique is relatively
1095 new, and as usage of CBCT increases, there may be potential for such injuries, particularly in
1096 cases of bad radiological protection practice.
1097
1098 Table 3.1. Estimates of threshold organ doses for tissue effects in adult human testes, ovaries, lens and
1099 bone marrow (Reproduced Table A.3.1. from ICRP, 2007b with updated information regarding eye
1100 lens and heart from ICRP 2012b).

Tissue and effect	Threshold Total dose in a single exposure (Gy)	Threshold Annual dose in the case of fractionated exposure (Gy/year)
Testes		
Temporary sterility	0.15	0.4
Permanent sterility	6.0	2.0
Sterility	3.0	>0.2
Lens		
Cataract (visual impairment)	0.5	
Bone marrow		
Depression of haematopoiesis	0.5	>0.4
Heart or brain		
Circulatory disease	0.5	

1101
1102 (48) Besides skin injuries, there have been recent reports of radiation effects on the lens of
1103 the eye, which is one of the most radiosensitive tissues in the body (ICRP, 2012b; Rehani et
1104 al., 2011). Radiation-induced cataracts have been demonstrated among workers involved in
1105 interventional procedures using x-rays (Vañó et al., 1998; ICRP, 2001) but not with CT or
1106 CBCT. However, an earlier study by Klein et al. (1993) and a more recent study by Yuan et al.
1107 (2013) has indicated that there may be elevated risk for damage to the lens of the eye in
1108 patients undergoing CT scans. Similar risks can be anticipated in patients undergoing CBCT,
1109 e.g. in neuroradiological interventions when the eye is exposed to the primary beam.
1110 Currently, there is a paucity of data and it is hard to judge the risk for patients. Caution is
1111 recommended where the primary beam irradiates the eye, and thus careful attention to
1112 optimisation is necessary.
1113 (49) In addition to patients, there are populations exposed to low doses in occupational
1114 settings. For some such groups, lens opacities have been documented, including workers in
1115 interventional suites (Rehani et al., 2011; Ciraj-Bjelac et al., 2010, 2012; Vano et al., 2010,
1116 2013); astronauts (Cucinotta et al., 2001; Rastegar et al., 2002), radiological
1117 technologists/radiographers (Chodick et al., 2008), atomic bomb survivors (Nakashima et al.,
1118 2006; Neriishi et al., 2007), and people affected by the Chernobyl accident (Day et al., 1995).
1119 (50) Recent epidemiological data suggest that tissue reactions can occur at threshold doses
1120 that are lower than previously considered (ICRP, 2012a,b). These reactions usually take a
1121 long time to manifest. For lens opacities, the threshold for damage is now considered to be as

1122   low as an absorbed dose of 0.5 Gy, whereas it was previously set at 2 Gy (depending upon
1123   exposure scenario). The absorbed dose threshold for circulatory disease has been chosen as
1124   0.5 Gy to the heart or brain, as a precautionary value. ICRP policy has been not to set any
1125   dose limits for patients. However, the current recommendation of the ICRP for occupational
1126   exposure in planned exposure situations is an equivalent dose limit for the lens of the eye of
1127   20 mSv/year, averaged over a defined 5-year period, with no single year exceeding 50 mSv
1128   (ICRP, 2012b). Occupational eye lens doses of a few μGy in CBCT have been reported in the
1129   literature. Eye lens doses for patients are a few mGy for dental and head and neck CBCT with
1130   direct exposure, but doses are much higher for interventional CBCT. Details regarding eye
1131   lens doses in CBCT for patient and personnel are available in Chapters 6 and 7.
1132
1133   <center>**3.3. Stochastic effects**</center>
1134
1135   (51) Cancer and heritable effects come into the category of stochastic effects. The
1136   probability of carcinogenic effects is much higher than heritable effects. This follows from
1137   the ICRP *Publication 103* (ICRP, 2007b) which states that the detriment-adjusted nominal
1138   risk coefficient or stochastic effects for the whole population after exposure to low doses of
1139   radiation is 5.5%/Sv for cancer and 0.2%/Sv for heritable effects. The latter is a theoretical
1140   risk for humans, as all documented cases of radiation-induced heritable effects come from
1141   observations in non-human species. Cases in humans have not been observed, even for
1142   survivors of Hiroshima and Nagasaki. Therefore, after careful review of many decades of
1143   literature, the ICRP has reduced the tissue-weighting factor for the gonads relating to the risk
1144   of genetic effects by more than half from 0.2 to 0.08 (ICRP, 2007b).
1145   (52) Major international organisations share the belief that the risk of developing cancer in
1146   patients exposed to radiation from CT scans is very low but appears to be more than
1147   hypothetical. Cancer risks are estimated on the basis of probability factors derived mainly
1148   from the survivors of Hiroshima and Nagasaki. There has been a tendency, in particular in CT,
1149   to use cancer risk estimates at individual patient level. This should be done with great care
1150   due to the large uncertainty of cancer risk estimates at low exposures. Furthermore, the ICRP
1151   recommends that "for the purposes of retrospective evaluation of radiation-related risks, such
1152   as in epidemiologic studies, it is appropriate to use sex- and age-specific data and calculate
1153   sex- and age-specific risks" (ICRP, 2007b).
1154
1155   <center>**3.4. Individual differences in radiosensitivity**</center>
1156
1157   (53) The individual differences in radiosensitivity are well known. Women and children
1158   are known to be more susceptible to radiation-induced cancer than men. For example, the
1159   lifetime attributable risk of lung cancer incidence for a 60-year-old woman exposed to 0.1 Gy
1160   is estimated to be 126% higher than that for a 60-year-old man exposed to the same dose
1161   (BEIR, 2006), and thus, gender considerations are important. A recent report from the United
1162   Nations Committee on Effects from Atomic Radiation (UNSCEAR) indicates that not all
1163   tissues in children are more sensitive to radiation (UNSCEAR, 2013). It is recommended that
1164   differences in radiosensitivity be taken into consideration during the justification process.
1165   Pre-existing autoimmune and connective tissue disorders, for reasons still not known, may
1166   predispose patients to the development of skin injuries of variable severity which cannot be
1167   predicted. Such disorders include scleroderma, systemic lupus erythematosus, and possibly
1168   rheumatoid arthritis. Genetic disorders that affect DNA repair, such as the defect in the ataxia
1169   telangiectasia mutated (ATM) gene responsible for ataxia telangiectasia, may be responsible

1170 for individual differences in radiosensitivity. Diabetes mellitus does not increase sensitivity to
1171 radiation, but does impair healing of radiation injuries (Balter et al., 2010).

1172

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1242
1243 **4. PRINCIPLES OF RADIOLOGICAL PROTECTION FOR PATIENTS AND**
1244 **WORKERS**
1245

1246 • **The ICRP recommends careful justification for each examination and procedure**
1247 **using CBCT.**

1248 • **The ICRP's concept of "as low as reasonably achievable" should be applied to**
1249 **achieve optimisation within DRLs.**

1250 • **Since many applications of CBCT involve patient doses similar to MDCT, the room**
1251 **layout and shielding requirements in such cases need to be similar to adequately**
1252 **protect workers.**

1253

1254   (54) The ICRP has been credited with development of the fundamental principles of
1255 radiological protection, which are justification, optimisation of protection and application of
1256 dose limits (ICRP, 2007). Dose limits are only applicable in radiation protection of workers
1257 and public; for patient protection, DRLs are used (ICRP, 2007b).

1258

1259   **4.1. Justification**

1260

1261   (55) The justification principle requires that the net benefit of radiation exposure be
1262 positive. According to ICRP, there are three levels of justification for the use of radiation in
1263 medicine.

1264 • At the first level, the use of radiation in medicine is acceptable when it results in more
1265   good than harm to the patient. It is now taken for granted that the use of x-rays in
1266   medicine is justified.
1267 • At the second level, a specified procedure with a specified objective is defined and
1268   justified (e.g. a CBCT examination for patients showing relevant symptoms, or a group of
1269   individuals at risk to a condition that can be detected and treated).
1270 • At the third level, the use of radiation in an individual patient should be justified (e.g. the
1271   particular CBCT application should be judged to do more good than harm to the
1272   *individual patient*).
1273   (56) According to ICRP *Publication 87* (ICRP, 2000), requests for a CT examination
1274 should be generated only by properly qualified medical or dental practitioners as defined by
1275 national educational and qualification systems. Justifying individual exposures should include
1276 verification that the information required is not already available from previous studies and
1277 that the proposed study is really going to answer the questions posed (ICRP, 2007a). The
1278 referring practitioners and imaging professionals should be skilled in the selection of, and
1279 indications for CT, CBCT and angiography, and possess adequate knowledge concerning
1280 alternative techniques. This training should also apply to non-imaging professionals who plan
1281 to use CBCT. Further aspects of training are provided in Chapter 8. The availability of
1282 resources and cost should also be considered in the justification process.
1283   (57) Justification of CBCT is a shared responsibility between the referring practitioner and
1284 the imaging professional. In the case of self-referral (e.g. practitioners in out-patient dental
1285 and ENT clinics) wherein the referring practitioner and the imaging professional are the same
1286 person, their responsibilities are combined within one person. Referring practitioners know
1287 their patients and their medical histories, but typically have little or even no knowledge about
1288 radiation doses, or the risks and limitations of diagnostic radiological examinations. On the

1289 other hand, imaging professionals have expertise regarding radiological examinations,
1290 including knowledge of alternate imaging examinations that can provide similar information
1291 with less radiation exposure to the patient; they, however, lack in-depth knowledge about the
1292 individual patient's condition. Consultation between imaging professionals and referring
1293 practitioners is essential to make the most of their combined knowledge. While such
1294 consultation has been emphasised before, practical constraints have made its implementation
1295 hard to realise in practice, and there is a need for exploration of tools to make this possible.
1296   (58) The ICRP has noted that there are many reports documenting lack of justification, in
1297 particular for CT examinations although not yet for CBCT (Fraser and Reed, 2013; Rehani
1298 and Frush, 2010). The ICRP recommends utilisation of modern technologies like use of
1299 clinical decision support system with electronic referral to improve justification.
1300
1301 ## 4.2. Optimisation
1302
1303   (59) Once an examination is justified, it must be optimised for that patient and the worker.
1304   (60) The primary role for optimisation of CBCT lies with the CBCT facility, and it should
1305 ensure that the examination is carried out with lowest radiation dose to the patient while
1306 obtaining the image quality required for the clinical purpose.
1307   (61) DRLs have been used to promote optimisation and have shown good results in many
1308 countries, particularly for CT applications. They were developed to identify examinations
1309 with doses above the 75th percentile in the dose distribution so that corrective actions could
1310 be taken. However, as expressed in the ICRP's concept of as low as reasonably achievable,
1311 they do not obviate the need for optimisation below the 75th percentile dose (Rehani, 2013).
1312 With modern technical equipment and optimised protocols, dose levels between the 25th and
1313 50th percentile are achievable (NCRP, 2012), so users should aim to optimise within DRLs
1314 (Rehani, 2013). The optimisation of patient protection in CBCT requires the application of
1315 examination-specific scan protocols tailored to patient age or size, region of imaging, and
1316 clinical indication. Protocols provided by the vendors of CT scanners should be evaluated for
1317 optimisation. DRLs are just one of the practical tools to promote the assessment of existing
1318 protocols. The ability to compare dose levels between CBCT facilities would facilitate the
1319 development of appropriate, new and improved protocols at each CBCT centre.
1320   (62) DRLs for CBCT procedures need to be established. To achieve this, doctors
1321 performing CBCT examinations should work closely with medical physicists.
1322
1323 ## 4.3. Requirements for imaging facilities
1324
1325   (63) Practice varies worldwide but should comply with requirements laid down by national
1326 authorities. Typically, each CBCT scanner should be registered with the appropriate database
1327 under the overall oversight of a national or designated authority. Frequently, during the
1328 process of registration and authorisation, an authority will examine the specifications of the
1329 machine and the size and shielding of the room where it is going to be used, ensuring that
1330 personnel and members of the public are sufficiently protected. The International
1331 Electrotechnical Commission (IEC, 2012) and the International Organization for
1332 Standardization provide international level safety requirements for x-ray machines. In many
1333 countries, national standards for x-ray machines are also available. These requirements are
1334 intended to protect workers and members of the public who may be exposed to radiation. The
1335 registration and authorisation process will also assess the availability of qualified staff. There
1336 are requirements for periodic quality control tests for constancy and performance evaluation.
1337 Acceptance tests and periodic quality control testing of CBCT equipment can provide

confidence in equipment safety and its ability to provide images of optimal image quality. Such periodic testing is also essential, because a malfunctioning machine may expose patients unnecessarily to radiation without any other overt signs. Nevertheless, whatever national requirements are, it is essential that they are followed in order to ensure that facility design and operation are safe for patients, workers, and the public.

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1368
1369   **5. ASSESSING PATIENT DOSES IN CBCT**
1370

1371   • **Traditional CT measurements with a 100-mm chamber are not sufficient for CBCT**
1372   **except for use as internal standard or reference. Dosimetry for CBCT is not yet**
1373   **standardised. Manufacturers should be encouraged to use consistent dose**
1374   **measurement units, and therefore, organisations responsible for establishing**
1375   **radiation units are encouraged to meet the challenge to avoid use of different**
1376   **quantities by manufacturers.**

1377   • **Equipment used for both fluoroscopy and CBCT needs to aggregate dose indices to**
1378   **individual patients during the entire procedure.**

1379   • **Measurement of dose variables in short phantoms does not provide an accurate**
1380   **indication of the overall dose. But, since determination of the complete rise-to-**
1381   **equilibrium dose requires very long phantoms of up to 600 mm, it is impractical to**
1382   **perform such measurements in the clinical environment. Therefore, manufacturers**
1383   **should measure and provide users with a full set of dosimetric data.**

1384   • **Manufacturers should also provide a subset of partial CT dose index (CTDI)**
1385   **measurements so that the complete rise-to-equilibrium curve measurements can be**
1386   **related to partial measurements that can be performed by users during acceptance**
1387   **testing of new equipment. While acceptance tests normally require both phantoms**
1388   **and free-in-air measurements, periodic measurement of $CTDI_{air}$ should be sufficient**
1389   **as long as free-in-air measurements remain stable with time.**

1390
1391   **5.1. Dosimetry in CBCT**
1392

1393   (64) CBCT utilises a wide x-ray beam for 3D imaging of a relatively large volume. Since
1394   the mid-1990s, the trend in MDCT has been towards an ever-increasing number of slices with
1395   a concomitant increase in x-ray beam width; the z-axis coverage of the high-end, wide-area
1396   MDCT scanners available today rivals that of CBCT. These developments have created a
1397   drive to update CT dosimetry methods so that they are more apropos wide-area detectors. As
1398   a result, some of the work from MDCT dosimetry, for which established measurement
1399   methods and phantoms already exist, can be translated to CBCT dosimetry. This chapter and
1400   its associated Annex A present the shortcomings of the standard narrow-beam MDCT
1401   formalism when it is directly applied to CBCT. Methods to overcome these problems are
1402   described in order to construct a comprehensive framework for CBCT dosimetry.
1403   (65) CT dosimetry has evolved around the concept of the CTDI. In order to connect the
1404   CTDI-like measurements with dose, volume CTDI ($CTDI_{vol}$) and dose length product (DLP)
1405   have been extensively used in clinical practice as relative patient dose indicators.
1406   (66) The limitation of this index for wider beams has led to new approaches in CT
1407   dosimetry, details of which are provided in Annex A. The CTDI paradigm is problematic
1408   when there is no helical scan or patient motion (as is the case with many CBCT scanners). In
1409   such cases, reported $CTDI_{vol}$ values will significantly overestimate the dose (Dixon and
1410   Boone, 2010a).
1411
1412   **5.2. Point of care scanning and physicians clinic based CBCT systems**
1413

1414   (67) Clinic-based systems include head and neck CBCT, breast CT (bCT) and dental
1415   CBCT. One of the main differences between dental and other clinic-based scanners (i.e. head
1416   and neck scanners) is the FOV, as head and neck scanners are capable of imaging larger
1417   volumes.

1418   (68) For dental systems, the SEDENTEXCT Consortium report (EC, 2012) discussed the
1419   use of dose/kerma-area product (DAP/KAP) as well as CTDI-like measurements. On the
1420   grounds that the conventional CTDI has drawbacks for dental CBCT use (due to wider beams
1421   and greater asymmetry of dose distribution in CBCT compared to MDCT), the consortium
1422   tried to define a single CBCT dose index (CBCT DI) (Pauwels, 2012). Further validation of
1423   possible indices is required, together with a way to translate dose indices' readings into
1424   patient doses. Araki et al. (2013) concluded that CBCT DI and KAP proposed by
1425   SEDENTEXCT could be used to establish DRLs in dental CBCT, but that the relationship of
1426   these indices to effective dose remains to be determined.

1427   (69) It has been suggested that if the manufacturer has provided a dose figure, then this
1428   quantity should be measured during commissioning. However, not all machines come with
1429   such initial measurements. The SEDENTEXCT Consortium proposes that if such
1430   measurements are not provided, the medical physicist should create a log of such readings in
1431   all clinically used settings so that the dentist may compare with national and international
1432   audit levels (EC, 2012).

1433   (70) Technically, the methods described above could also be applied to other clinic-based
1434   systems including systems for dental and head and neck imaging and possibly bCT. However,
1435   there is currently no standardisation in the measurements for such units. This highlights more
1436   vividly that the issue of standardisation in CBCT dosimetry remains largely unresolved.

1437

1438   **5.3. C-arm CBCT systems**

1439

1440   (71) C-arm CBCT systems are incapable of performing a full rotation around the patient
1441   couch. Some systems, however, can rotate only 180° plus the beam angle (Fahrig et al., 2006),
1442   which results in a non-uniform axial dose deposition to the patient/phantom. In a phantom,
1443   the maximum dose occurs at the central plane intersecting the z-axis at z = 0, on the side of
1444   the phantom closest to the x-ray tube. In the ideal case in which the heel effect is absent, the
1445   maximum dose would occur on the bisector of the rotation angle. When the heel effect is
1446   present, the maximum dose occurs near the bisector.

1447   (72) For C-arm CBCT systems, Fahrig et al. (2006) proposed a metric representing the
1448   average dose to the phantom central plane, following a similar averaging to that applied in
1449   calculation of the weighted CTDI ($CTDI_w$).

1450

1451   **5.4. A unified approach to CT dosimetry**

1452

1453   (73) The International Commission on Radiation Units and Measurements (ICRU) in their
1454   report No. 87 (ICRU, 2013) has reviewed a considerable body of work in order to propose a
1455   method for CT dosimetry that compensates for the shortcomings of current CTDI-based CT
1456   dosimetry methods. In addition, earlier work by Dixon and Boone (2010b) provided a unified
1457   formalism for dose measurements on machines capable of helical scanning (e.g. MDCTs) as
1458   well as on those that only acquire axial images (which is the case with most CBCTs). A set of
1459   metrics and the use of a new polyethylene 600 mm long phantom are proposed. The
1460   mathematical foundation for the method is beyond the scope of this publication, but the
1461   method is briefly discussed in Annex A.

(74) The physical interpretation of the rise to equilibrium curve presented in Annex A is that the scan and the phantom need to be long enough so that the asymptote tails of the profiles are reached. The longer the scan, the closer H(L) approaches to unity. This representation shows that the dose to the central CT slice in a scan increases with scan length, demonstrating the relatively low efficiency of short scans for collecting the actual dose; this efficiency increases with longer scans.

5.5. Tracking and reporting of radiation dose

(75) New challenges emerge with systems being used for both fluoroscopy and tomography (CBCT). Currently, there is no standardised way to assess the aggregate radiation dose to a patient during a single procedure. This situation needs to be addressed, and these imaging systems should provide a means of not only comparing but also consolidating doses from both the fluoroscopy and CT components of a procedure. Furthermore, tracking and reporting of the radiation dose for a single patient should be facilitated.

5.6. Epilogue

(76) The unified CT dosimetry method proposed by ICRU (2013) has the potential to standardise CBCT dosimetry. Nevertheless, the value of CTDI-based measurements should not be underestimated. Although CTDI has limitations, it has been evaluated on many systems over the years and provides important comparisons in output for CT scanners from different manufacturers and ages. Moreover, coefficients for patient dose estimations based on the $CTDI_{vol}$ are already available.

5.7. References

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1511
1512 **6. OPTIMISATION OF PATIENT AND WORKER DOSES IN CBCT**
1513

1514 • **Optimisation of both patient and worker doses, particularly when workers have to**
1515 **be near the machine, is important wherein monitoring of doses become an essential**
1516 **tool. Recording, reporting and tracking of radiation dose for a single patient**
1517 **should be made possible.**

1518 • **Low dose protocols may be sufficient to answer diagnostic questions focussed on**
1519 **high-contrast structures, such as lung, bones, dental scans (teeth and maxillofacial),**
1520 **ENT scans (paranasal sinuses, skull, temporal bone), interventional material, and**
1521 **contrast-enhanced vessels (angiographic interventions).**

1522 • **Protocols with higher dose should only be selected if visualisation of soft-tissue**
1523 **structures, such as intracranial haemorrhage, soft-tissue tumours, and abscesses, is**
1524 **the primary focus.**

1525 • **Most interventional and intra-procedural C-arm CBCT systems can scan an**
1526 **angular range spanning 180 to 240 degrees + the cone angle of the x-ray beam. The**
1527 **radiosensitive organs, such as thyroid, eyes, female breast and gonads, should be**
1528 **on the "detector side" of the arc, whenever possible.**

1529 • **Clinical need permitting, every effort should be made by users to ensure that the**
1530 **volume of interest is fully incorporated in the FOV provided by the CBCT**
1531 **scanners while radiosensitive organs are placed outside the FOV.**

1532 • **Post-processing tools such as "thick slice reformats" allow averaging of adjacent**
1533 **slices to lower image noise. This may be sufficient for answering certain diagnostic**
1534 **questions and evaluation of soft-tissue structures.**

1535 • **The aim of CBCT should be to answer a specific diagnostic or intra-operative**
1536 **question *vis-à-vis* other imaging modalities and not to obtain image quality that**
1537 **rivals MDCT. The decision by the referring practitioner to utilise CBCT should be**
1538 **made in consultation with imaging professional.**

1539 • **The user must understand the consequences of scan protocol selection not only in**
1540 **terms of image quality, but also in terms of applied dose. This is especially**
1541 **important for CBCT, where such information may be entirely (and sometimes,**
1542 **ambiguously) encoded in the protocol name.**

1543 • **There is a need to provide checks and balances, for example dose check alerts**
1544 **implemented in CT in recent years, to avoid high patient doses as compared to**
1545 **locally defined reference values.**

1546 • **Methods which provide reliable estimates of eye dose under practical situations**
1547 **should be established and utilised.**

1548
1549 **6.1. Introduction**
1550

1551 (77) CBCT scanners are highly engineered machines and dose optimisation is a
1552 multifactorial problem. The imparted radiation dose may vary by several orders of magnitude
1553 between different scan modes and use scenarios. Clinical use of CBCT requires insight into
1554 the various trade-offs in order to maximise patient benefit and minimise risk. It is essential to

1555 understand various technological factors and scan parameters that influence dose. Knowledge
1556 of MDCT alone is not sufficient in this endeavour as CBCT scanner systems are significantly
1557 different in their mode of operation from MDCT scanners. For example, while spiral scanning
1558 is the norm with MDCT, nearly all CBCT imaging is done using a single axial scan. In
1559 addition, several special conditions exist that do not apply to MDCT scanners (e.g. the
1560 restriction on the FOV of a typical CBCT scanner). It is therefore essential to involve a
1561 medical physicist or another suitably qualified expert early on in optimisation, as well as the
1562 audit of patient and occupational dose levels, particularly for high dose procedures.
1563
1564 <div align="center">**6.2. Factors influencing dose to the patient**</div>
1565
1566 **6.2.1. Equipment dependent factors**
1567
1568 *Knowing your equipment*
1569 (78) It is important that users understand how their equipment functions, because each
1570 CBCT scanner has some unique features, such as the application domain, gantry design, and
1571 detector configurations. The complexity of modern equipment necessitates a thorough
1572 understanding of the various scan modes, parameter settings, and dose optimisation strategies.
1573 This section deals with equipment features that have bearing on radiation dose, and the next
1574 section is devoted to operator actions required to achieve optimal radiation protection in
1575 clinical scans.
1576
1577 *Collimation*
1578 (79) In MDCT, the region of interest is usually prescribed on one and sometimes two
1579 orthogonal scan projection radiographs (also known as antero-posterior (AP) and lateral
1580 (LAT) scout views or topograms); the scanner then helically or axially covers this scan FOV
1581 and reconstructs tomographic slices. Similar AP and LAT projection views may also be
1582 acquired in CBCT scanning; however, the entire FOV usually fits within a single circular
1583 trajectory of the scanner and helical scanning is not used in most applications. Although most
1584 of the time the x-ray beam will not extend beyond detector dimensions in situations where the
1585 detector is movable, a portion of the beam may fall outside the detector margins. Care should
1586 be taken to collimate the x-ray beam so that it falls entirely within the detector margins;
1587 automatic means for delimiting the collimation window to the detector size may or may not
1588 exist, depending on the particular scanner make and model. Any radiation outside the detector
1589 constitutes unnecessary radiation to the patient. The beam should be further collimated to
1590 limit its z-extent to the FOV. The source-to-detector distance determines the maximum lateral
1591 extent of the FOV that can be scanned and should be appropriately adjusted depending on the
1592 anatomy under consideration. It should be noted that the scatter noise in the projection data
1593 increases approximately linearly with the area of the irradiated field. In general, the x-ray
1594 beam should be tightly collimated as it not only lowers the x-ray dose, but simultaneously
1595 decreases scatter thereby improving image quality.
1596 (80) A poorly collimated primary beam, if it is outside the patient, may significantly
1597 increase the occupational dose, as well as the patient dose. It is also desirable to exclude from
1598 the scan FOV any adjacent sensitive organs that do not need to be imaged to address the
1599 clinical question at hand. The x-ray beam should be tightly collimated to the scan FOV. As a
1600 CBCT scan cannot be extended in the same way as an MDCT one, caution must be exercised
1601 to ensure that the volume of interest is fully incorporated in the FOV provided by the CBCT
1602 scanner.
1603

1604   *Collimation along the z-axis*
1605   (81) Many CBCT scanners provide a means for the user to collimate the beam. Collimation
1606 along the z-axis to achieve as narrow a beam as possible to fulfil the clinical purpose will
1607 both reduce the patient dose and improve the image quality. Use of the thinnest possible
1608 collimation (2.3 cm) instead of the full field (19 cm) improves contrast to noise ratio.
1609   (82) Free-in-air geometric efficiency is a means of quantifying over-beaming, i.e. the
1610 proportion of radiation falling outside the detector margins (Berris et al., 2013). In CBCT
1611 scanners, the x-ray beam is usually fully intercepted by the receptor, so the free-in-air
1612 geometric efficiency should be 100%, and over-beaming should not occur. Furthermore,
1613 over-scanning (aka over-ranging) which is required at either end of helical scans to provide
1614 additional data for image reconstruction, is not needed for axial CBCT scans (Tzedakis et al.,
1615 2005).
1616   (83) An effect that always occurs in CBCT is that parts of the irradiated volume are hit by
1617 radiation, but are not fully contained in 180° of projections. Images of these regions, shown in
1618 Fig. 6.1., cannot be reconstructed or can only be partially reconstructed. The region that
1619 cannot be reconstructed broadens as the cone angle increases (Grimmer et al., 2009).
1620
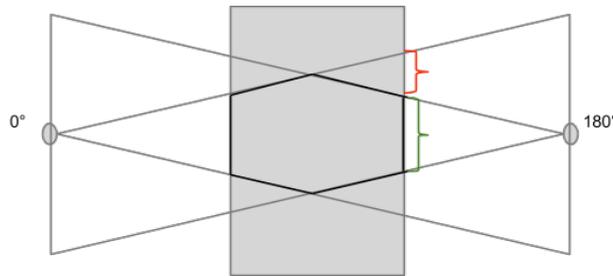
1621
1622 Fig. 6.1. In CBCT, only within the region in the hexagon that is marked with the green
1623 parenthesis is data available from 180° projections. However, a part of the irradiated volume
1624 (red parenthesis) cannot be reconstructed (or only with reduced image quality), because there
1625 is no data from all 180° of projections available. The size of this area depends on the
1626 geometry of the scanner (qualitative depiction). (permissions required)
1627
1628   *Dose distribution within the scan field of view along the z-axis*
1629   (84) Ideal CT scanner systems should irradiate the examined volume along the z-axis with
1630 a homogenous dose that should fall off rapidly outside the examined volume. In some CBCT
1631 systems, the dose distribution is different, and the central slices receive larger amounts of
1632 radiation (Gupta et al., 2006). Wherever possible, radiosensitive organs should be placed
1633 outside the irradiated volume, which is normally wider than the FOV, provided the clinical
1634 requirements of the procedure permit.
1635
1636   *Dose distribution in case of volume-of-interest scanning*
1637   (85) In certain situations, only a small volume such as a couple of teeth and the adjacent
1638 bone may be of clinical interest. Some CBCT scanners provide a very narrow beam
1639 collimation with a relatively small detector. A large part of the irradiated volume will be out
1640 of the primary x-ray beam at most angular projection positions. In general, a scan volume that
1641 is delimited in the x-y-direction to a small portion of a larger body part results in truncation
1642 artefacts. However, small volume CBCT of high-contrast structures such as bones and teeth,
1643 when used in conjunction with an artefact reduction algorithm, may well give clinically
1644 acceptable images. For example, a truncation artefact arising from a limited FOV may not
1645 affect assessment of a transpedicular screw. This must not be confused with retrospective,
1646 selective reconstruction of a certain region of interest inside a larger scanned volume (See

1647 Table. 6.1.). The dose distribution outside the volume of interest is very different in the two
1648 scanning modes. Therefore, the user should verify whether volume-of-interest scanning is
1649 applicable in a certain situation.
1650
1651 Table 6.1. Volume of interest scanning versus standard scanning: Volume of interest scanning is a
1652 great method to reduce the radiation exposure of in-plane structures, if imaging conditions allow it
1653 (high-contrast structures). It must not be confused with standard scanning for region of interest
1654 reconstruction.

	Irradiated volume from all directions (from all angular positions)	Reconstructed volume	Radiation exposure	Applications
Volume-of-interest scanning	Limited to cylindrical volume of interest	Limited to cylindrical volume of interest	Only volume-of-interest receives full dose	Mostly dental imaging, maxillofacial imaging and most interventional C-arm setups when body trunk is scanned
Standard scanning	Large cross section	Anywhere within body diameter, full body diameter or parts of full cross section	Whole body diameter receives full dose	All other

1655
1656 *Type of detector*
1657 (86) Most currently available CBCT systems use a digital FPD. State-of-the-art digital
1658 FPDs are offered at several gains and effective dynamic range settings. In general, the
1659 dynamic range of digital FPDs is narrower than for MDCT detectors, resulting in poorer soft-
1660 tissue contrast for CBCT scanners. The afterglow of the caesium iodide (CsI) scintillators
1661 used in FPDs limits the maximum image frame rate that can be obtained from these detectors.
1662 Typically, 30 FPS can be obtained at the full FOV; a narrower FOV can provide a faster
1663 frame rate of 100 to 120 FPS (Gupta et al., 2008). Slow frame acquisition rate is the main
1664 reason for the relatively high acquisition times of CBCT systems; the fastest clinically
1665 available CBCT, as of 2013, has an acquisition time of 5 seconds as compared with 80
1666 milliseconds for a dual source MDCT system (Orth et al., 2008). Parameters such as pixel
1667 size and scintillation crystal thickness are usually selected based on target application (e.g.
1668 maxillofacial imaging or C-arm angiography), and the end user has no control over their
1669 selection. Currently, there is no detector technology being employed that should be strictly
1670 avoided from a radiation protection standpoint.
1671 (87) A minority of CBCT systems still uses CCD cameras coupled with x-ray image
1672 intensifiers (XRII). The convex input screen and image distortion of image intensifier systems
1673 result in non-uniform image quality across the output image. In addition, light and electron
1674 scattering within the image intensifier limits the contrast resolution of the reconstructed slices.
1675 CBCT systems typically have an 8 to 10-bit dynamic range and can only support a very
1676 coarse level of tissue differentiation.
1677
1678 *Detector quantum efficiency*
1679 (88) The DQE is a widely used metric that describes the dose efficiency of an x-ray
1680 detector. Without going into details, it measures the quality of the image produced by the

1681 detector from a given dose or fluence to the detector. Intuitively, it captures how well a
1682 detector translates the signal incident on it into an image, relative to an ideal detector.
1683 Specifically, it is the square of the ratio of input and output SNR of a detector. For example, a
1684 detector that reduces the SNR by 50% has a DQE of 0.25. The ideal detector would have a
1685 DQE of one and would translate all incident x-ray photons into image information. DQE is
1686 normally given as a function of spatial frequency and correlates image quality with incident
1687 x-ray dose at a detector level.
1688 (89) Current caesium iodide hydrogenated amorphous silicon (CsI-aSi:H) FPDs have
1689 DQEs in the range of 0.6–0.7, which are lower than that of MDCT detector systems (Gupta et
1690 al., 2006). This is a fundamental limitation, which is beyond the control of the user, and
1691 means that for the same input radiation, the CBCT images will be noisier than MDCT images.
1692
1693 *Filtration*
1694 (90) A bowtie filter in the imaging chain hardens and attenuates the x-ray beam, reduces
1695 the scatter-to-primary ratio, and reduces the x-ray fluence heterogeneity at the detector.
1696 Bowtie filters decrease the scatter contribution from the object periphery in MDCT imaging
1697 (Orth et al., 2008). Ning et al. (2000) have shown that the quantity [SNR^2/entrance exposure]
1698 decreases when kVp increases for a flat-panel-based CBCT system. This means that there is a
1699 trade-off between decreased scatter from the object periphery (when the bowtie filter is on)
1700 and improved detector efficiency from the "softer" beam (without bowtie filter) (Orth et al.,
1701 2008). Use of bowtie filter is standard in MDCT. In CBCT, a bowtie filter is not used
1702 commonly, but its use is increasing. Other configurations such as half bowtie filters that
1703 enable large area coverage have also been used (Wen et al., 2007). The presence of the filter
1704 can reduce patient dose, especially at the patient periphery, and can improve tomographic
1705 image quality by improving uniformity, CT number accuracy, and contrast to noise ratio. One
1706 potential disadvantage, however, is the decrease in detector efficiency due to beam hardening
1707 (Mail et al., 2009). In general, a bowtie filter should be used when imaging a wide FOV
1708 where the anatomy under consideration occupies only a small central portion. Assessment of
1709 spinal hardware would be one example application. Special care must be taken if the bowtie
1710 filter is removable; workers can forget to mount the bowtie filter prior to imaging resulting in
1711 additional dose to the patient.
1712
1713 *Anti-scatter grid*
1714 (91) An anti-scatter grid is placed between patient and detector, and consists of lead septa
1715 that are oriented along lines projecting radially outwards from the focal spot. This geometry
1716 allows the primary beam to reach the detector while the off-axis radiation is absorbed. As
1717 such, an anti-scatter grid in front of the flat panel can prevent the scatter generated by the
1718 patient from reaching the detector. The leaves reduce the effective detector area to a small
1719 degree. The geometry of the anti-scatter grid, which determines its selectivity and its rejection
1720 efficiency, is optimised for the scanner and application. Anti-scatter grids are highly sensitive
1721 to the source-to-detector distance; if the latter can be varied, or a choice of anti-scatter grids is
1722 provided, it is essential to match these two parameters.
1723 (92) The efficiency of anti-scatter grids for scatter suppression and image quality
1724 improvement has been assessed for CBCT. Although the presence of a grid did not seem to
1725 improve the SNR in relation to applied radiation dose (Schafer et al., 2012), a significant
1726 decrease in cupping artefacts was observed (Kyriakou and Kalender, 2007). However, in
1727 certain high scatter conditions, the grid could lead to a reduction in dose of up to 50%
1728 (Kyriakou and Kalender, 2007).

1729   (93) The anti-scatter grid, if available, is usually a fixed hardware parameter that is
1730 optimised for a certain application and a specific geometry. Typically, the end user has little
1731 influence on the geometry of the anti-scatter grid. However, if a choice of different grids and
1732 geometric distances is provided, it is essential that the two are matched for the system to
1733 function properly.
1734
1735 *Scatter correction algorithm*
1736   (94) Scatter intensity has a broad angular distribution around the image of the scattering
1737 object. One can think of the projection image obtained by the detector as a 2D smeared image
1738 of the object that includes both the primary and the scatter radiation. At any point that can
1739 receive both the primary and scatter photons, these two components may be difficult to
1740 separate. However, in areas that are shielded from the primary beam by the collimator, the
1741 scattered component is observable because of broad distribution of the scatter. An assessment
1742 of this can be used to estimate the amount of scatter in the rest of the image. By assuming a
1743 scattering function, the scatter profile throughout the image can be estimated. This can then
1744 be subtracted from the measured signal to compute the contribution from the primary
1745 signal. If a particular CBCT scanner provides a set of steps for computing the scatter function,
1746 that protocol should be strictly followed. Besides vendor-implemented algorithms, the user
1747 has little influence over the scatter correction algorithms.
1748
1749 *Data correction algorithms*
1750   (95) Multiple correction algorithms are typically applied to the raw projection data, before
1751 it can be reconstructed into a 3D stack. The following is a partial list of data conditioning
1752 algorithms typically employed to compensate for system imperfections: (1) offset subtraction;
1753 (2) afterglow correction; (3) adaptive filter mask; (4) normalisation; (5) theta correction; (6)
1754 cross-talk difference correction; (7) air calibration; (8) Gordon scaling; (9) beam hardening
1755 correction; and (10) detector z-gain non-uniformity correction. These corrections tend to be
1756 vendor specific and the end-user has no control over them.
1757
1758 **6.2.2. Operator dependent factors**
1759
1760 *Reduced arc scanning*
1761   (96) Many CBCT systems are capable of reconstruction from less than 360 degree angular
1762 acquisitions. In general, a coverage of 180 degrees plus the cone angle is sufficient for
1763 tomographic reconstruction. This gives the operator considerable flexibility in selectivity, so
1764 allowing reduction of patient exposure. For example, an appropriate choice of starting and
1765 stopping angle can be used to limit projection images of a patient's head to posterior angles,
1766 reducing the dose to the lens of the eyes (Kyriakou et al., 2008) (Fig. 6.2.). Daly et al. (2006)
1767 observed a 5-fold decrease in eye dose when 3D images were generated using a C-arm half-
1768 cycle (178°) rotation performed with the x-ray tube posterior to the skull rather than anterior.
1769 Another example where this is used is in CBCT imaging of the breast, where the imaging
1770 angles can be chosen to limit unnecessary exposure to the heart and lungs. These manoeuvres
1771 typically have no appreciable effect on the image quality in the central portions of the scan.
1772 Selecting an appropriate angular span for the scan arc, a parameter that has a direct impact on
1773 the dose distribution, is a user-selectable parameter. The user should select the scan arc so
1774 that radiosensitive organs are on the detector side of the imaging chain.
1775   (97) Dental CBCT differs regarding the use of a reduced arc. Firstly, the start- and
1776 endpoints of a 180° rotation cannot be selected by the user, with the detector typically being
1777 at the anterior side of the patient. However, simulations and phantom studies have pointed out

1778 that patient dose may be lower when the tube is at the anterior side, although differences were
1779 10% or lower (Morant et al., 2013; Zhang et al., 2013; Pauwels et al., 2012). This can be
1780 explained by the anterior placement of FOVs for dental examinations, which results in several
1781 radiosensitive organs being posterior to the centre of rotation (e.g. parotid salivary glands).
1782 More evidence is needed before a definitive recommendation can be made to manufacturers.
1783
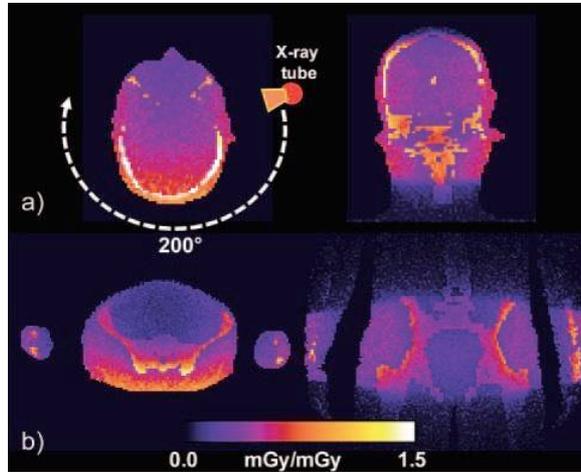
1784
1785 Fig. 6.2. In contrast to MDCT scanning, CBCT scanning is mostly performed with a half scan
1786 angle (180°+cone-angle). This gives the position of the scan angle a significant influence on
1787 the dose distribution within the patient. (Kyriakou et al., 2008). (permissions required)
1788

1789 *Setting of kVp and mAs*
1790   (98) The parameters that determine x-ray beam flux and energy spectrum (i.e. the mA and
1791 kVp settings) should be kept as low as possible without compromising the image quality and
1792 clinical utility of the scan. The kVp and mA are the main user selectable variables that
1793 determine the overall dose to the patient. If all other parameters are held constant, the
1794 radiation dose is directly proportional to the applied mAs (tube current × the duration of the
1795 scan rotation), and this parameter significantly influences the noise in the image. As long as
1796 the detector is not saturated, there is a direct relationship between the level of image quality
1797 and increasing mAs. The dependence of the radiation dose and image quality on the kVp
1798 setting is more complex. Higher-energy photons result in less interaction with tissue; they
1799 give poorer contrast between tissues, but a larger number of photons pass through the tissue
1800 and reach the detector to form the image. The right kVp and mAs setting depends heavily on
1801 the anatomy being scanned, whether or not a contrast medium was used, and also depend on
1802 several design factors such as filter systems, frame rate, and detector type. Therefore, it is
1803 difficult to provide absolute guidelines. All commercial CBCT scanners come with a
1804 manufacturer recommended protocol for each application. The best advice to the user is to
1805 start with this protocol, and working in conjunction with a medical physicist or another
1806 domain expert, to adapt it to the local conditions. One should also monitor publications and
1807 guidelines dedicated to the special scanner setup or type of examination.
1808

1809 *Automatic exposure control*
1810   (99) AEC in CBCT systems adapts the radiation exposure to obtain a desired level of
1811 image quality and adjusts the dose to that needed for the specific body part of the patient.
1812 Similar to MDCT, AEC modulates the tube current according to patient attenuation in a given
1813 angular direction. Usually, AEC is implemented as a feedback loop that controls the x-ray

1814 source based on feedback from the detector. Reductions in dose by 20-40% through the use of
1815 AEC systems have been reported (McCollough, 2005; He et al., 2010).
1816   (100)   Many CBCT systems do not employ AEC, using instead a fixed tube current setting
1817 for the entire scan. The utility of tube current modulation is reduced in CBCT due to the wide
1818 z-axis coverage. Also, the demand for AEC is less stringent when scanning the head as
1819 compared to other parts of the body. The requirements and demands on the AEC are still
1820 evolving, and general guidelines are difficult to formulate. More details on the patient-
1821 specific factors involved in the potential application of AEC can be found in section 6.2.3.
1822
1823 *Scan modes: number of projections*
1824   (101)   In contrast to MDCT scanning, where the user is unable to influence the number of
1825 projections explicitly, this parameter is often directly selectable in CBCT. The most
1826 commonly used detectors in CBCT systems are much slower in readout and require a wait-
1827 time after each projection in order to account for the afterglow of the scintillator. The dose
1828 delivered in each scan is also limited because of the number of photons that can be collected
1829 by each projection without overexposing the detector. Optimisation of the scan time using a
1830 tight control over each exposure is much more critical in CBCT than in MDCT. These
1831 considerations limit the range of dwell time and dose in each projection. By controlling the
1832 number of projections, for example or, by changing the total scan time, one can control the
1833 dose for a scan protocol: increasing the number of projections proportionately increases the
1834 applied radiation dose. In CBCT, the number of projections, together with the associated
1835 changes in the total scan time, provides a trade-off between image quality and the delivered
1836 dose that is directly influenced by user-selected parameters.
1837
1838 *Scan modes: binning and spatial resolution*
1839   (102)   The detector elements in angiographic C-arm CBCT systems, in contrast to MDCT
1840 detector systems, are much smaller in order to provide the necessary spatial resolution for
1841 fluoroscopy and angiography modes. For example, a common FPD for C-arm systems offers
1842 a native pixel size of 154 μm in a 1,920 × 2,480 matrix. The time to readout such a large
1843 matrix, coupled with the afterglow of the CsI scintillators, limits the maximum frame rate
1844 achievable on such a detector. The frame rate of a CBCT detector can be as much as 1 to 2
1845 orders of magnitude lower than that in MDCT. Low readout frame rate accounts for the
1846 relatively high acquisition times of CBCT systems. For example, the fastest available clinical
1847 CBCT, as of 2013, had an acquisition time of few seconds as compared to 0.08 milliseconds
1848 for a dual source MDCT system (Orth et al., 2008).
1849   (103)   While one cannot do much about the afterglow or after-lag of the scintillator, the size
1850 of the image matrix that needs to be readout can be decreased to make the image transfer
1851 faster. A set of binning modes is provided to accomplish this. Each binning mode combines
1852 neighbouring detector rows and columns in order to reduce the matrix size and the readout
1853 time. Typical binning modes involve a 2×2 and 3×3 area, thereby reducing the data to be
1854 streamed out by a factor of 4 and 9, respectively. Despite this averaging, the spatial resolution
1855 of CBCT is higher than that in MDCT and is often above the demands of the clinical
1856 application. Since the image noise, spatial resolution and radiation dose are interrelated, the
1857 user must decide on the acceptable image quality and the spatial resolution. This choice, in
1858 turn, determines the radiation dose. The user should not be tempted to reduce the image-noise
1859 – e.g. by increasing the tube current or increasing the number of projections using modes such
1860 as the "high-quality scan modes" offered on some systems – to reach a noise level that is
1861 comparable to that of MDCT. The dose penalty associated with these scans can be much
1862 higher than would be warranted by the clinical question at hand (Blaickner and Neuwirth,

47

1863 2013). Post-processing techniques, such as slice averaging, thick multi-planar reformation,
1864 use of a softer reconstruction kernel, are preferable when trading off among competing
1865 metrics such as image noise, low contrast resolution, spatial resolution and radiation dose.
1866
1867 *Scan modes: predefined scan protocols*
1868 (104) The use of an organ-specific protocol (e.g. "routine head") or a clinical indication-
1869 specific protocol (e.g. "appendicitis protocol") is an established practice in MDCT. In routine
1870 clinical care, vast libraries of such scan protocols are available. Similar to MDCT, many
1871 CBCT systems also provide predefined scan protocols that encapsulate detector settings,
1872 reconstruction kernels and other scanner parameters. In CBCT, however, the usage is less
1873 well established with many protocols named suggestively with prefixes such as "low" or
1874 "high-quality", the latter unflatteringly implying that the base protocol might not provide
1875 appropriate image quality in certain situations (see Table 6.2.). Generally, the naming of the
1876 scan protocols refers to the well-known and, within limits, physically fixed trade-off between
1877 image quality parameters and radiation dose. "High-quality" scan protocols usually provide
1878 "better" image quality at "higher" radiation dose. These simple prefixes often belie the
1879 magnitude of the change that occurs: a "high-quality" protocol may entail a 6-10 fold increase
1880 in radiation dose as compared to a low or standard quality protocol. In CBCT, the selection of
1881 the scan mode or scan protocol is one of the most significant factors influencing radiation
1882 dose (Kyriakou et al., 2008). A low-dose scan protocol may be sufficient for high-contrast
1883 structures, such as bones, teeth, kidney stones and contrast-enhanced blood vessels. The
1884 manufacturers are beginning to provide scan protocols that are named for the diagnostic
1885 challenge they are trying to address (e.g. "bone", "kidney stone", "rule out intracranial
1886 haemorrhage" or "skull base" protocol). There may be a dedicated section for paediatric
1887 protocols. These have special significance when the imaging system does not have an AEC
1888 (e.g. in most dental CBCT scanners) to account for the lower diameter of children´s body
1889 parts.
1890 (105) The user-interface for CBCT scanners also deserves a special mention. The checks
1891 and balances that are routine in MDCT may be missing in CBCT scanners. For example, two
1892 vastly different but similarly named protocols may be adjacent to each other on the user-
1893 interface, or a single mouse click may cause a 10-fold change in the delivered dose. This is in
1894 sharp contrast to MDCT where such a big increase in radiation requires several purposeful
1895 manipulations of scan parameters and concomitant confirmation to affect the change. The
1896 user must understand the consequences of scan protocol selection not only in terms of image
1897 quality, but also in terms of applied dose. This is especially important for CBCT, where such
1898 information may be entirely (and sometimes, ambiguously) encoded in the protocol name.
1899 There has been considerable variability in lexicon used in imaging that creates difficulty in
1900 dose registry. The Commission recommends standardisation of lexicon used in imaging
1901 protocols.
1902
1903 Table 6.2. Overview of available scanning protocols, applications and typical protocol names.
1904 Protocols that are only a single click away from each other have vastly different dose consequences. In
1905 addition to patient positioning and selection of the scanning arc, appropriate protocol selection is the
1906 most significant user determined factor for radiation dose calculation.

Protocol dose	Protocol spatial resolution	No of projections	Regions	Clinical indication	Names (examples)
Low	Low	Low	Abdomen,	Rule out kidney stone,	"-", "low-

			Thorax	assess position of instrument/implants, Treatment planning	quality", "low-dose"
Medium	High	Low/Medium	Skull/Bones	Maxillofacial imaging, dental imaging, assess bone structures, arterial contrast media angiography	"dental", "bone", "high-resolution"
High	High	High	Abdomen, Head	Assess soft-tissue structures, intracranial haemorrhage, venous contrast media angiography	"+", "CT-angiography", "high-quality"

1907
1908 *Scan modes: Partial panel*
1909   (106)   In order to expedite readout of the panel, the detector control electronics generally
1910 allows readout of partial panel: an arbitrary number of only the central rows may be read out
1911 as needed. While most systems have built-in hardware features that ensure effective use of the
1912 beam, it is essential, from a radiation protection point of view, that the x-ray beam is
1913 appropriately collimated to irradiate only that portion of the detector that is being read out.
1914
1915 *Keep unnecessary body parts out of the x-ray beam*
1916   (107)   It is good practice to limit the radiation field to the body parts that must be imaged.
1917 Inclusion of unnecessary body parts not only has dose consequences, but also may
1918 significantly increase image artefacts. Many CBCTs have only a limited scan-FOV, with a
1919 diameter lower than the body region that is being examined. Positioning of arms or legs
1920 outside the irradiated area can significantly reduce the level of artefacts and therefore increase
1921 the image quality without increasing unnecessary radiation dose.
1922
1923 *Making judicious use of CBCT acquisitions during a procedure*
1924   (108)   CBCT imaging can quickly provide 3D images intra-operatively with minimal effort
1925 on the part of the interventionalist or surgeon. These datasets are useful since they relieve the
1926 operators from the effort of trying to distinguish overlapping structures in 2D fluoroscopy
1927 images. They can also save dose by replacing multiple DSA runs in different C-arm
1928 angulations with a single CBCT run. It has been shown that the 3D acquisition provides
1929 valuable clinical information and limits the need for 2D imaging: hence, CBCT can also
1930 lower the dose in one procedure. Given this facility, and the ease with which 3D images can
1931 be acquired, operators may be tempted to overuse the 3D imaging features of their equipment.
1932 Even though CBCT has the potential to decrease dose in comparison to fluoroscopy and
1933 MDCT, this effect could be cancelled by overuse of volumetric acquisition with C-arm and
1934 other intra-operative CBCT machines. 3D data must be judiciously acquired for purposeful
1935 clinical problem-solving only when fluoroscopy is insufficient for the task at hand.
1936
1937 *Bismuth shielding*
1938   (109)   Bismuth shielding for the eyes, thyroid, breast or other organs in CBCT should be
1939 used with caution. However, reduced arc scanning will be more effective (section 6.2.2.) and
1940 such shielding must not be used in conjunction with this. Bismuth shielding can be effective
1941 in certain situations if placed in a manner that does not interfere with the AEC system of the

CBCT scanner. If the shield is positioned after the AEC has adjusted tube current to be used, then this may be beneficial provided the image is not excessively degraded by the presence of the shields in the FOV (AAPM, 2012). If the bismuth shield is placed before selection of the AEC, its effect may be totally negated by the increased current from the AEC.

Reconstruction algorithms

(110) In a standard CBCT reconstruction algorithm such as the modified Feldkamp-Davis-Kress (FDK) algorithm, the noise level is proportional to the applied radiation and tube current. However, image filtering, compressed sensing, and iterative reconstruction algorithms, which are becoming increasingly popular in MDCT, have the potential to disrupt this direct relationship between the applied dose and image quality. At the present time, such novel reconstruction algorithms are not widely available for CBCT scanners, and it is not possible to provide specific guidelines on how they should be used in practice. In many circumstances, the application of these specialised algorithms is not universal. Instead, a user-selectable mixing parameter is provided. This percentage factor determines the level to which the output of the specialised reconstruction algorithm should be incorporated and added to the output of the traditional algorithm. The exact setting for this mixing factor will depend on the algorithm and the acceptable image quality, and will have to evolve with experience.

6.2.3. Patient-specific factors

Thickness of the body part in the beam

(111) In response to the varying thickness of the anatomy, most CBCT machines adjust radiation exposure automatically through an AEC. This electronic system has a sensor that detects how much signal is being produced at the image receptor, and adjusts the x-ray generator to increase or decrease exposure factors (typically tube current and in many cases tube voltage) so that each projection image is of a consistent quality. When a thicker body part is in the beam, or a thicker patient is being imaged (compared with a thinner patient), the machine will automatically increase the exposure. The result is a similar image quality but an increase in the entrance dose.

(112) In MDCT, the AEC is able to vary the tube current both in the angular as well as the longitudinal or z-direction. As a result of the angular variation, the dose in the AP direction is lower than that in the lateral direction for any fixed, user-selected image quality parameters. The z-axis adaptation of the dose controls the mA value in the superior-inferior direction, resulting in a higher dose to the abdomen and pelvis as compared to the chest. In CBCT, since most acquisitions are performed in an axial rather than a helical mode, the angular variation of tube current is more important.

(113) Some CBCT systems lack an AEC. These systems operate under the assumption that the patient size does not vary significantly in the angular direction. This assumption can be true for dental and head-and-neck applications, but should be further investigated.

*Children in CBCT*

(114) For any given exposure settings (same tube settings, collimation, amount of projections, etc.), a thinner patient will receive a higher dose (which is energy deposited per mass) than a larger patient, even though the larger patient absorbs a greater fraction of the radiation (AAPM, 2011b). This is because the lower attenuation in a thinner body results in a smaller range in dose through the body tissues for the smaller patient (e.g. a paediatric patient). This may also sometimes be true even when the exposure factors are adjusted for body size or are controlled by an AEC. In general, especially for large patients, a greater

1991 fraction of the x-ray beam is absorbed in the more superficial portions of the anatomy being
1992 imaged. In other words, the skin dose is much higher than the central dose. For thinner
1993 patients, this dose gradient is smaller, which implies that the dose is high throughout the
1994 entire body. Figs. 6.3. and 6.4. illustrate the absorbed radiation dose as a function of the
1995 patient's body habitus and size when an AEC compensates for variations in body size. Thus,
1996 it is important to pay particular attention to optimising radiation protection for children to
1997 ensure that exposure factors are not higher than necessary.
1998
1999

A   B   C
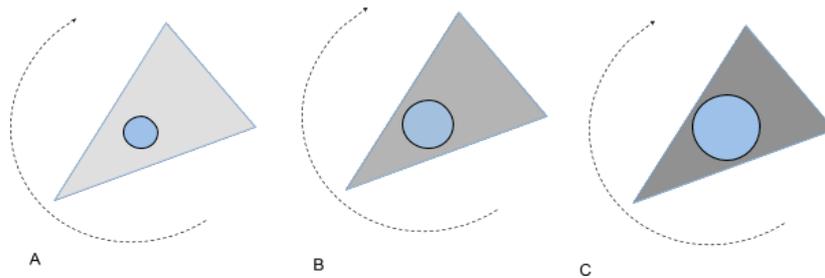
2000
2001 Fig. 6.3. Qualitative illustration of the effects of an AEC on patient exposure. The AEC keeps
2002 the image quality at a given level and adjusts for variations in patient size. The impact of
2003 patient size on the radiation dose with the AEC is shown, while panel A shows the smallest
2004 patient diameter, panel C is the largest patient diameter, and panel B is in between them.
2005 Radiation exposure is indicated by grey level of the radiation fan. The bigger the patient, the
2006 higher the applied radiation exposure. (permissions required)
2007

The tube output increases at angular positions that irradiate a larger patient diameter

The tube output increases at angular positions that irradiate a larger patient diameter (derived from an actual thorax scan)
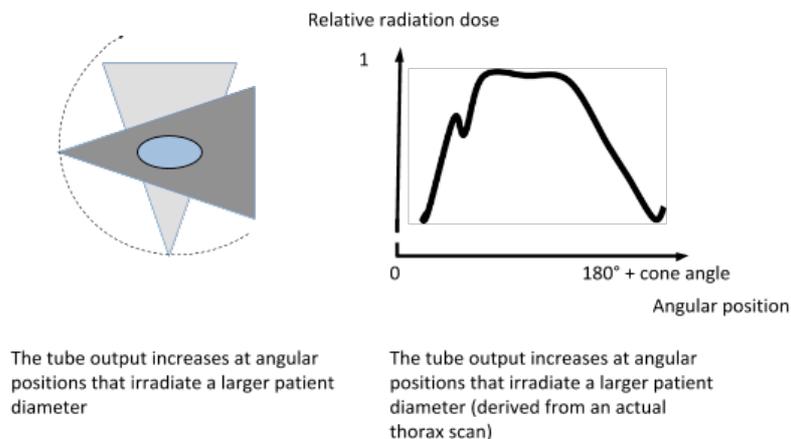
2008
2009 Fig. 6.4. The effects of the variation in the patient diameter in-plane is demonstrated using
2010 AEC. At angles where the larger patient diameter is greater, the exposure is increased. The
2011 diagram is an example derived from an actual torso scan (as provided by Rolf Kueres).
2012 (permissions required)
2013

Monitoring of patient dose indices

2015 (115) Unfortunately, the field of patient dose monitoring in CBCT lags behind that in
2016 MDCT. There is a lack of standardisation in dosimetry methods for CBCT; different
2017 manufacturers have provided different ways of measuring and reporting dose in CBCT and
2018 these are not universally adopted. It is hoped that if the recommendations of ICRU Report 87
2019 (ICRU, 2012) are adopted by manufacturers and clinicians, there is a good possibility that
2020 dosimetry in CBCT will be standardised and will provide more coherent patient dose data in
2021 the future. Means to estimate and report patient dose will require a collaborative effort
2022 between the manufactures of CBCT equipment and the regulatory bodies. Methods for storing

2023 patient dose indices and dose reports in Picture Archiving and Communication Systems
2024 (PACS) also have to evolve as the use of CBCT becomes more prevalent.

2025   (116)   In view of recent cases of skin injuries to patients in CT examinations, there is a
2026 need to provide checks and balances to avoid over exposures through alerts and prospectively
2027 control patient dose in comparison to locally defined reference values (Cadet, 2010; NEMA,
2028 2013; AAPM, 2011; RPOP, 2010). Manufacturers need to incorporate suitable features to
2029 facilitate this.

2030
2031 **6.2.4. Factors influencing dose to worker**
2032

2033   (117)   Occupational radiation exposure is expected to be small in the case of clinic-based
2034 CBCT systems. While using a C-arm or other CBCT devices in an interventional suite or
2035 operating theatre, physicians, technologists and other workers can protect themselves by
2036 using shielding devices. As required under national regulations in most countries, radiation
2037 workers must comply with regular individual dose monitoring requirements for managing
2038 radiation exposure and keep a comprehensive dose record. Further, unless necessary, worker
2039 should move outside the fluoroscopy room, when CBCT acquisition is taking place.

2040   (118)   In one study, the unshielded CBCT exposure at 35 cm distance from the operating
2041 table, measured over a 60-second scan, was found to be 0.26 mSv (Daly et al., 2006). Schulz
2042 et al. (2012) measured eye dose ranging from 28.0 to 79.3 μSv for CBCT hepatic arterial
2043 embolisation and biliary tube placement procedures. The primary source of radiation is the x-
2044 ray tube, and ideally, the patient alone should be exposed to the primary x-ray beam.
2045 Radiation scattered from the patient, parts of the equipment, and the patient table - the so-
2046 called 'secondary radiation' or 'scatter radiation' - is the main source of radiation exposure to
2047 the worker. A useful rule of thumb is that radiation dose rates are higher on the side of the
2048 patient closest to the x-ray tube. Distance is also an important factor, and when permitted in
2049 the clinical situation, workers should increase their distance from the x-ray source and the
2050 patient. Automatic injectors should be used, as far as possible, if contrast medium injection is
2051 necessary.

2052
2053 *Shielding: Lead apron*
2054   (119)   Clinical staff taking part in diagnostic and interventional procedures using C-arms
2055 for fluoroscopy or CBCT imaging wears protective aprons containing lead (sometimes also
2056 lined with additional x-ray absorbent materials) to shield tissues and organs from scattered x-
2057 rays (NCRP, 1995). Transmission through these aprons will depend on the energies of the x-
2058 rays and the lead-equivalent thickness of the aprons. If the attenuation of scattered radiation is
2059 assumed to be equal to that of the primary (incident) beam, this provides a margin of safety
2060 (NCRP, 2005).
2061   (120)   All workers present in the room during a CBCT scan must wear a lead apron, as it is
2062 the most essential component of personal shielding in an x-ray room. It should be noted that
2063 the level of protection afforded by the lead apron depends on the x-ray energy, which is a
2064 function of the voltage applied across the x-ray tube (kV). The thicker the part of the patient's
2065 body falling in the x-ray beam, the higher the kV set by the fluoroscopic system. Higher kV
2066 x-ray photons have greater penetrative power, implying that a greater lead thickness is needed
2067 to provide the necessary attenuation.
2068   (121)   For procedures performed on thinner patients, particularly children, an apron of 0.25-
2069 mm lead equivalence will suffice. However, for thicker patients and with a heavy workload, a
2070 0.35-mm lead apron may be more suitable. The wrap-around aprons of 0.25-mm lead
2071 equivalence are ideal; these have a thickness of 0.25 mm at the back and 0.5 mm at the front.

Two-piece skirt-type aprons help to distribute the weight, and due to their overlap in front of the abdomen, they provide a 1-mm shielding, e.g. at the level of the uterus. Heavy aprons can pose a problem for workers who have to wear them for long periods of time. There are reports of back injuries due to the weight of lead aprons among workers who wear them for many years (NCRP, 2010). Some newer aprons are lightweight while maintaining lead equivalence, and have been designed to distribute the weight through straps and shoulder flaps.

Ceiling-suspended shielding

(122) Ceiling-suspended screens that contain lead impregnated in plastic or glass are very common in interventional radiology and cardiology suites. However, they are not usually used in operating theatres. Shielding screens are very effective as they have lead equivalences of 0.5 mm or more and can reduce x-ray intensity by more than 90%. Practical problems make the use of radiation shielding screens for occupational protection more difficult but not impossible in operating theatres. Manufacturers should develop shielding screens that can be used for occupational protection without hindering the clinical task. There is a need for more than one screen to effectively provide protection to other personnel in the operating theatre in addition to the main operator.

Mounted shielding

(123) These can be table-mounted lead rubber flaps or lead glass screens mounted on mobile pedestals. Lead rubber flaps are very common in most interventional radiology and cardiology suites, but are rarely seen in operating theatres; nevertheless, their use should be promoted. Manufacturers are encouraged to develop detachable shielding flaps to suit practices in operating theatres. Lead rubber flaps, normally impregnated with the equivalent of 0.5 mm lead, should be used as they provide effective attenuation.

Room shielding

(124) Room shielding requirements for CBCT systems used in dental and maxillofacial imaging range from 0.5- to 1.5-mm lead equivalent, depending on the scanner's specifications for scattered radiation dose and its workload (EC, 2012). In most cases, the image receptor intercepts the entire primary beam, as in most fluoroscopic units and MDCT scanners. The room shielding is for scattered radiation, as is the case with a conventional CT scanner (Sutton et al., 2012). However, for any type of CBCT machine, the shielding should be designed to keep doses to workers and the public as low as reasonably achievable and of course below the existing dose limits that apply in various settings.

Lead glasses

(125) Various types of leaded glass eyewear are commonly available, although they are heavier than the common glass eyewear. These include eyeglasses that can be ordered with corrective lenses for individuals who normally wear eyeglasses. There are also eye shields that can be clipped onto the spectacles of workers, and full-face shields that also function as splash guards. Leaded eyewear should either have side shields to reduce the radiation coming from the sides or be of a wrap-around design with angled lenses. The use of protective devices for the eyes as well as for the body is recommended.

Individual protection and monitoring

(126) The principles of radiological protection of workers from ionising radiation are discussed in *Publication 75* (ICRP, 1997) and reiterated in Paragraph 113 of *Publication 105*

2120 (ICRP, 2007b). In this section, practical points pertaining to those who need to be monitored
2121 and what protective actions should be taken are discussed.

2122 (127) Individual monitoring of workers exposed to ionising radiation using film dosimeters,
2123 thermoluminescent dosimeters (TLDs), optically stimulated luminescence (OSL) badges, or
2124 other appropriate devices is used to verify the effectiveness of radiation protection practices
2125 in the workplace. The advice of a radiological protection expert/medical physicist should be
2126 sought to determine which method is most appropriate. An individual monitoring programme
2127 for external radiation exposure is intended to provide information about the optimisation of
2128 protection and to demonstrate that the worker's exposure has not exceeded any dose limit or
2129 the level anticipated for the given activities (IAEA, 1999). As an effective component of a
2130 programme to maintain exposures as low as reasonably achievable, it is also used to detect
2131 changes in the workplace and identify working practices that minimise dose (NCRP, 2000;
2132 IAEA, 2004). In 1990, the Commission recommended a dose limit for workers of 20
2133 mSv/year (averaged over a defined 5-year period; 100 mSv in 5 years) and other limits as
2134 given in Table 3.1.; these limits were retained in the 2007 Recommendations (ICRP, 1991,
2135 2007a). However, all reasonable efforts to reduce doses to the lowest possible levels should
2136 be used.

2137 (128) The Commission recommended that interventional radiology departments develop a
2138 policy that staff should wear two dosimeters (ICRP, 2000). A single dosimeter worn under the
2139 lead apron will yield a reasonable estimate of effective dose for most instances. Wearing an
2140 additional dosimeter at collar level above the lead apron will provide an indication of the
2141 thyroid dose (if unprotected) and other parts like head and the lens of the eye. In view of
2142 increasing reports of radiation-induced cataracts in those involved in interventional
2143 procedures, monitoring the dose to the eye is important (Ciraj-Bjelac et al., 2010; Vañó et al.,
2144 2010). Recently, eye lens dosimetry has become an active research area. Many studies have
2145 been performed to determine which personal dose equivalent quantity is appropriate, and how
2146 it can be used for monitoring the dose to the lens of the eye, and to develop dosimeters to
2147 measure dose to the lens of the eye (Domienik et al., 2011). The Commission recommends
2148 that methods which provide reliable estimates of eye dose under practical situations should be
2149 established and utilised.

2150 (129) A risk-based approach to occupational radiation monitoring should be adopted to
2151 avoid unnecessary monitoring of all workers. There is a need to raise awareness of the
2152 requirement to use a dosimeter at all times, as there are many examples of infrequent use in
2153 practice.

2154 (130) The lack of use or irregular use of personal dosimeters is still one of the main
2155 problems in many hospitals (Miller et al., 2010; Padovani et al., 2011). The protection service
2156 should provide specialist advice and arrange any necessary monitoring provisions (ICRP,
2157 2007a). In cases where individual monitoring is inappropriate, inadequate, or not feasible, the
2158 occupational exposure of the worker should be assessed on the basis of the results of
2159 monitoring the workplace and information about the locations and durations of exposure of
2160 the worker (IAEA, 1996). In addition to individual monitoring, it is recommended that
2161 indirect methods using passive or electronic dosimeters (e.g. dosimeters attached to the C-arm
2162 device) should be used in these installations to enable the estimation of occupational doses to
2163 professionals who do not use their personal dosimeters regularly. Active dosimeters are an
2164 asset in the education and practice of radiation protection.

2165
2166 **6.3. Limitations of CBCT**
2167

6.3.1. Detector dynamic range and reduced contrast resolution

(131)   Compared to the detector system used in MDCT scanners, the FPDs have a lower dynamic range and lower DQE. For example, the contrast resolution of FPD-based CBCT is about 10 HU, which is inferior to the 1-3 HU available on MDCT. Therefore, applications that require imaging of low-contrast structures (e.g. grey-white matter differentiation in a head CT) will perform poorly on a CBCT scanner as compared with MDCT.

6.3.2. Scatter

(132)   The large FOV of these scanners implies that the entire volume generates the scatter radiation. Since an anti-scatter grid, which would further decrease the efficiency of the imaging chain, is not used typically, scatter can significantly degrade image quality.

6.3.3. Temporal resolution

(133)   FPDs usually employ CsI as the scintillator. CsI is a slow scintillator and suffers from afterglow (i.e. a ghost of the old image is seen in the new image at fast frame rates). As a result, after each projection, sufficient time must be allowed to elapse before the next projection is recorded.

6.3.4. Artefacts

(134)   CBCT images in general suffer from more or less the same types of artefact that are seen in MDCT, but to different degrees. A summary of MDCT artefacts has been provided by Barret et al. (2004). Metal and windmill artefacts are generally reduced in CBCT compared to MDCT, particularly for high-density metals (Pauwels et al., 2013). Motion artefacts, on the other hand, are more prevalent in CBCT imaging.

(135)   In MDCT, a smaller number of slices, typically 4 to 64, although up to 320 slices in some scanners, are acquired in each rotation as the patient is translated through the gantry. Therefore, any patient motion affects only those slices that were being acquired during the motion. In CBCT, the entire dataset is constructed from projections acquired in one rotation. Therefore, any motion, however short-lived, affects the entire volumetric dataset. The rotation speed of CBCT compared to MDCT is about 10-20 times slower, hence CBCT is much more sensitive to motion artefacts.

6.3.5. Hounsfield Unit consistency

(136)   The HU system is based on the linear attenuation coefficient of water. All CT scanners present clinical images in this system for consistency across vendors and scanner models. The daily calibration of MDCT scanners incorporates scanning of a water cylinder for HU calibration and beam hardening correction. CBCT scanners typically lack detailed radiometric calibration, and the generated HU values are more variable than those from an MDCT scanner. In contrast to MDCT, truncation of the body outlines and drawbacks of the reconstruction algorithm, lead to cupping artefacts. When scanning a homogeneous water phantom, the HU units are not uniform over the entire cross section, but decline towards the edges (Kyriakou et al., 2011).

2216 **6.3.6. Geometric distortion**
2217

2218 (137) Depending on the type of gantry used, a CBCT scanner is more prone to geometric
2219 distortions than MDCT. For example, when a C-arm is used as a CBCT scanner, the weight
2220 of the gantry may deform the unit, so that the isocentre of the imaging chain is not as well-
2221 defined. This will degrade the image quality. In addition, flexible alignment of many of the
2222 CBCT gantries necessitates a collision-avoidance system that may increase the complexity of
2223 a scan.
2224

2225 **6.4. Future developments**
2226

2227 (138) Several technical developments in the field of CBCT are expected to enable
2228 interesting new features that will affect image quality and imparted radiation. Since these
2229 features are only at an early stage of development, and mature implementations are
2230 unavailable in the scanner systems currently in use, only general guidance about their efficacy
2231 and application can be given at this point in time.
2232

2233 **6.4.1. Novel scan trajectories**
2234

2235 (139) For tomographic reconstruction, projective data from a rotation of at least 180° plus
2236 cone angle are necessary. This requirement imposes several constraints on the design and
2237 operation of CBCT in practice. For example, C-arm systems need to have a large clearance in
2238 the operating room to complete the scan trajectory, and lack of space may limit the utility of
2239 certain scan modes of the C-arm CBCT in practice. Novel scanning trajectories, such as
2240 eccentric rotation and/or parallel shifting of the imaging chain, may relieve some of these
2241 constraints and be useful in extending the scan FOV. These newer, non-traditional scan
2242 trajectories lead to a much more complex distribution of the applied dose in the examined
2243 volume. Currently, only one commercial robot CBCT system uses these alternative
2244 trajectories. However, the dose estimation systems are not designed to handle such systems.
2245 In the future, radiation protection measurements will have to account for these non-traditional
2246 trajectories and factor in the associated non-uniform dose deposition.
2247

2248 **6.4.2. Advanced methods for exposure control**
2249

2250 (140) AEC is a means to adapt the scan parameters to an individual patient's anatomy and
2251 its variations. Usually, the AEC is provided by a feedback loop between the radiation
2252 measured at the detector side and the x-ray tube exposure settings. In its simplest form, the
2253 tube current is varied so as to keep the total radiation measured at the detector constant. This
2254 compensatory mechanism can fail when the patient size increases beyond a certain
2255 point. After that point, for a given kV setting, the x-ray tube may not be able to deliver a
2256 further increase in mA without overheating or causing damage to the x-ray tube anode.
2257 Sometimes, in order to accommodate such large variations in photon flux, when current
2258 modulation alone is not able to meet the demand, in CBCT, x-ray tube voltage setting is also
2259 changed by the AEC. This practice is rare in MDCT, and in fact, interferes with the fidelity of
2260 HU calibration, but it is common practice in fluoroscopy. In order to make this practice
2261 workable for CBCT, most manufacturers use experimentally measured correlation graphs
2262 between measured x-ray photons and x-ray tube settings (current as well as voltage).
2263 (141) If tube voltage would be changed during a scan, inconsistencies in the measured CT-
2264 values with respect to the Hounsfield scale definition have to be taken into account and

2265 corrected. AEC with tube current as well as voltage variations make actual patient dose
2266 estimations from tube parameters and phantom experiments very complex. As this practice
2267 becomes more prevalent, further research will be needed in this area of dose measurement
2268 practice in order to account for this non-traditional use of the AEC systems.
2269
2270 **6.4.3. Novel reconstruction algorithms and compressed sensing**
2271
2272 (142) Analytical reconstruction algorithms, such as the filtered back projection, have been
2273 the mainstay for MDCT. These algorithms provide a single pass solution that is available on
2274 nearly all CT scanners. Even though they are generally fast and provide good image quality,
2275 they tend to be prone to noise and artefacts. In the past decade, a new class of iterative
2276 reconstruction algorithms has been introduced for MDCT by various vendors. Instead of
2277 using an analytical approach, these algorithms attempt to minimise the error between the
2278 projections and the reconstructed slices. Typically, 1-30 iterations are required for the
2279 solution to converge. These algorithms generally provide better image quality, and are more
2280 robust in minimising noise and artefacts. Their main drawback, besides their complexity, is
2281 their slow computational speed. They are generally associated with increased image
2282 resolution, decreased radiation dose, and metal artefact reduction. They can also be used for
2283 region-of-interest reconstruction.
2284 (143) Currently, a non-iterative, modified FDK algorithm is the industry standard for
2285 image reconstruction in CBCT. Similar to the reconstruction algorithms for the MDCT
2286 systems, where the use of iterative reconstruction algorithms is now gaining in popularity, a
2287 shift in CBCT reconstruction from a modified FDK to an iterative technique is expected.
2288 These reconstruction methods have the ability to incorporate prior knowledge in the form of
2289 radiation and scatter distribution, as well as knowledge of the anatomy. They also minimise
2290 the error between the projections and the reconstructed image in a global sense. These
2291 features would be advantageous for CBCT, since it is often performed in situations where
2292 repetitive scanning of the same anatomical region is necessary, for example, to observe the
2293 evolution of a contrast bolus through the vasculature and the tissue. Another example of
2294 repetitive scanning would be angiographic interventions to deploy interventional devices such
2295 as aneurysm coils and confirm its position. Often, changes in the successive 3D volumes are
2296 relatively minor. Iterative algorithms can accommodate these requirements more readily and
2297 so minimise the number of projections required for 3D or 4D reconstruction.
2298 (144) In order to reconstruct a volume of interest or a slice, a minimum number of data
2299 points are needed, in a strict mathematical sense, for the reconstruction task. If the dose per
2300 projection is fixed, this minimum number of projections determines the overall patient dose.
2301 If certain assumptions can be made about the object, and the requirement that projection
2302 images be equally spaced is relaxed, an image can be reconstructed under conditions which
2303 contravene the Nyquist–Shannon limit (i.e. the theoretical minimal sampling rate required for
2304 reconstruction). These methods, which are generally called compressed sensing, can reduce
2305 the dose by reducing the number of input projections required for reconstruction. Sparse
2306 angular sensing where projections are acquired only from certain angular direction, is one
2307 method for reducing dose using compressed sensing.
2308 (145) Both iterative reconstruction techniques and compressed sensing are in their infancy
2309 in CBCT. However, these novel techniques are expected to greatly impact image quality and
2310 the associated radiation dose in CBCT in the future. The user has to be aware that long
2311 established relationships between radiation dose and image quality may undergo fundamental
2312 changes with the use of novel, iterative reconstruction algorithms.
2313

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2452
7. RADIATION DOSE MANAGEMENT IN SPECIFIC APPLICATIONS OF CBCT
2454
- **The user of CBCT in interventions can significantly influence the radiation dose imparted to the patient by judiciously using a "low-image-quality or low dose" vs. a "high-image-quality or high dose" scan.**

- **In radiotherapy, justified use of CBCT has potential at different stages of therapy such as: pre-treatment verification of patient position and target volume localisation, evaluation of non-rigid misalignments, such as flexion of the spine or anatomic changes in soft-tissue, and during or after treatment to verify that the patient position has remained stable throughout the procedure. Low-dose CBCT protocols should be used for pre-treatment alignment of bony structures.**

- **Many machines were initially only capable of fluoroscopy, but can now additionally perform CBCT. Because of the improved clinical information in CBCT, and its ability to remove overlying structures, the user may be tempted to over utilise the CBCT mode. Users should judiciously use CBCT mode.**

- **In orthopaedics, justified use of CBCT can help in assessing the position of fractures and implants with respect to the bony anatomy, especially in situations where fluoroscopy alone is insufficient and thus help in patient dose management.**

- **In urology, low-dose CBCT protocols should be used when imaging high-contrast structures, such as calcified kidney stones.**

- **Dental CBCT scans should be justified, considering 2D radiography as an alternative, and optimised through the use of small FOVs and application- and patient-specific exposure factors.**

7.1. Introduction

(146) CBCT is used in a multitude of clinical applications. To maximise the practical utility of this report, this chapter is organised according to different clinical application domains that use CBCT rather than design considerations as they tend to be very similar across different applications. For example, a C-arm system used in interventional radiology (neuro, non-vascular, vascular) differs only marginally, if at all, from that used in orthopaedics or urology. However, application-specific radiation varies considerably across these domains, primarily because of patient-related and use-related factors. At the end of each section, practical tips on the use of the CBCT are provided that are germane to that application domain.

(147) This chapter also cites and summarises various published studies that provide typical range of CBCT dose values for each clinical application domain. Absolute dose values are provided and may be used by a practitioner as a reasonable starting point.

(148) It should be stressed that disparate methods have been used in the literature to measure and quantify dose. Many manufacturers provide concise dose values for their machines under varying scanning conditions and protocols. Often such data are required for the regulatory approval process. It is recommended that the user consult these documents and dose databases. But even such documents that have been submitted to regulatory agencies for licensing, suffer from a lack of standardisation in dose measurement techniques and units.

(149) The drawing of conclusions from the published studies and vendor documents, especially when absolute dose values are compared, should be done with care, keeping in

2498 mind the limitations of such comparisons because of variations in the measurement
2499 methodology. It is expected that future published literature on CBCT will use dose
2500 measurement guidelines similar to those provided in Chapter 5. Such standardised and
2501 consistent dose figures will enable direct comparisons among different machines, protocols,
2502 and imaging practices. In parallel, standardisation of DICOM dose reporting for CBCT is
2503 needed in order to enable retrospective retrieval and review of patient exposure from stored
2504 PACS images.

7.2. CBCT in radiotherapy

(150) The primary role of CBCT in radiation therapy is pre-treatment verification of
patient position and target volume localisation. In the most common pattern of workflow, a
patient lies on the treatment couch, is positioned approximately for treatment using wall-
mounted lasers, and then precise positioning is based on CBCT imaging. In addition to
correcting the position of the patient, the images are examined for non-rigid misalignments,
such as flexion of the spine or anatomic changes in soft-tissue. CBCT imaging is also
sometimes acquired during or after treatment to verify that the patient position has remained
stable throughout the procedure. CBCT can also be used in treatment simulation, prior to the
beginning of a course of treatment.

(151) Most radiation therapy centres use gantry-mounted kV CBCT, with an x-ray tube as
the source and amorphous silicon flat-panel imagers as detectors (Jaffray et al., 1999).
Typical energies are between 80 and 125 kVp, with typical absorbed doses within the
imaging volume between 1 and 40 mGy. A less-common modality is MV CBCT, using the
treatment accelerator as an x-ray source and a portal imaging FPD (Pouliot et al., 2005). MV
CBCT generally uses energies of up to 6 MV, with typical absorbed doses between 20 and
100 mGy. Compared with kV CBCT, the images produced with MV CBCT generally have
lower soft-tissue contrast, due to the lack of photoelectric absorption at higher photon
energies. However, these systems do have some advantages, including better geometric
alignment of imaging and treatment isocentres, and better imaging for large patients or
patients with metallic prostheses.

(152) The choice of imaging technique is based on the treatment site and therapy goals.
For cranial or head and neck targets, the treatment site is well accounted for by alignment of
bony anatomy. Therefore, a low-dose CBCT technique is appropriate. Similarly, when the
treatment target can be aligned using implanted fiducial markers, a low dose technique is
warranted. In these cases, accurate positioning with CBCT can be performed with absorbed
doses less than 10 mGy. Accurate positioning in the pelvis and abdomen, however, may
require differentiation of soft tissue boundaries. In these cases, the number of photons used
for imaging should be increased and may require an imaging dose between 10 and 40 mGy.

(153) The overall absorbed doses to tissues of a patient within the field imaged by CBCT
are small compared to the prescribed treatment dose. However, the treatment dose is localised
to the disease site, whereas the CBCT imaging dose is spread across the entire imaging
volume. When compared to other pre-treatment imaging modalities, CBCT can provide better
setup accuracy with equal or lower dose than MV port films (Korreman et al., 2010), but uses
more dose than orthogonal planar kV x-ray imaging (Kry et al., 2005) or non-ionising setup
methods such as optical imaging or ultrasound. Furthermore, one must keep in mind that the
primary radiation fields produce Compton scattered x-rays which deposit dose in the
neighbourhood around the treatment site. The magnitude of the scattered dose depends upon
the distance from the treatment field, and ranges from about 0.05% to 0.5% of the dose at d_{max}.

2546 The radiation dose at d_{max} is defined as 100% and it decreases as the penetration through
2547 tissue increases, the decrease primarily coming from the energy absorbed within the tissue.
2548
2549 Table 7.1. Doses in CBCT procedures in radiotherapy. Listed values are for a single CBCT acquisition
2550 and should be multiplied by the number of CBCT scans performed to compute the total dose.

Procedure	Reported values	Measurement technique	Reference
MV CBCT head and neck	150 mGy	Absorbed dose to isocentre	Pouliot et al., 2005
MV CBCT head and neck pelvis	60-73 mGy 99-121 mGy	TLD measurements on central plane	Gayou et al., 2007
kV CBCT head and neck chest pelvis	1 – 17 mGy 11 – 18 mGy 24 – 54 mGy	$CTDI_w$	Song et al., 2008
kV CBCT head and neck pelvis	36.6 mGy 29.4 mGy	CB $CTDI_w$	Cheng et al., 2011
kV CBCT head and neck chest pelvis	2.1 – 10.3 mSv 5.2 – 23.6 mSv 4.9 – 22.7 mSv	TLD measurements at 26 locations in anthropomorphic phantom	Kan et al., 2008
kV CBCT head and neck pelvis	1.1 ± 0.5 mGy 36 ± 12 mGy	TLD measurements at 22 locations in anthropomorphic phantom	Stock et al., 2012
kV CBCT chest	Spinal cord: 8-22 mGy Left lung: 12-29 mGy Right lung: 16-40 mGy Heart: 17-30 mGy Body:12-31 mGy	Absorbed doses from Monte Carlo simulation	Spezi et al., 2012
kV CBCT head and neck	Spinal cord: 1.3-1.7 mGy Mandible: 4.5-8.3 mGy Right parotid: 0.3-2.7 mGy Left parotid:0.5-2.7 mGy Left eye: 0.1-1.8 mGy Right eye: 0.1-1.8 mGy Oral cavity: 1.7-3.8 mGy Body: 1.0-2.3 mGy Brainstem: 0.3-1.5 mGy Larynx:2.6-2.8 mGy	Absorbed doses from Monte Carlo simulation	Spezi et al., 2012
kV CBCT pelvis	Rectum dose: 11-21 mGy Left femoral Head: 20-47 mGy Right femoral head: 25-62	Absorbed doses from Monte Carlo simulation	Spezi et al., 2012

	mGy Body: 11-33 mGy		
kV CBCT thorax MVCT thorax (non CBCT)	0.9-20.6 mGy 0.3-9.1 mGy	TLD thorax phantom measurements in breast, heart, lung, abdomen, sternum, rib, thyroid	Shah et al., 2012
kV CBCT pelvis MV CBCT pelvis	17.9-50.6 mGy 0.9-8.0 mGy	TLD pelvis phantom measurements in prostate, bladder, rectum, sigmoid, left femoral head, right femoral head	Shah et al., 2012
kV CBCT pelvis MV CBCT pelvis kV CBCT head MV CBCT head TomoTherapy pelvis	25-40 mGy 40-80 mGy 1-7 mGy 30-50 mGy 13 mGy	IMRT phantom measurements with radio-photoluminescent glass dosimeter	Kouno et al., 2013
kV CBCT Head & Neck Chest Pelvis	19 mGy 51 mGy 167 mGy	Measurement of primary doses at the centre of custom-made phantom using a glass dosimeter	Kim et al., 2013
KV CBCT Pelvis Head & Neck	0.2-6.7 mGy 0.03-0.7 mGy	Measurement of secondary doses (20-50 cm from isocentre) measured on custom-made phantom using a glass dosimeter	Kim et al., 2013
kV CBCT thorax full-rotation scan limited arc scan	5.00 ± 0.30 mSv 2.44 ± 0.21 mSv 1.23 ± 0.25 mSv 1.17 ± 0.30 mSv	Measurements of dose to organs performed with radiochromic film	Alvarado et al., 2013

2551

7.2.1. Accounting for imaging dose in radiotherapy

2553

(154)   When x-ray imaging is used in a radiotherapy setting, the patient receives radiation from both imaging and therapy. CBCT imaging, especially when employed daily, causes additional accumulated dose which should be considered in the context of the patient's treatment. For this reason, the use of daily CBCT imaging should be evaluated for each patient for sparing sensitive organs that have low thresholds for deterministic effects, and for paediatric patients who have a higher sensitivity to radiation.

(155)   With first generation linac-mounted kV CBCT systems, imaging doses can account for 2% or more of the prescribed target dose (Amer, 2007; Ding, 2008, 2009). However, the current trend is toward dose reduction, and second generation systems have achieved significant dose savings in kV-CBCT (Ding and Munro, 2013). When the imaging dose

2564 constitutes a significant fraction of the prescription dose (ICRU Report 83), it should be
2565 reflected in the patient's prescription dose. For example, the prescription dose can be
2566 adjusted to include the imaging dose. A more advanced accounting procedure is to perform
2567 patient-specific CBCT dose calculation in the Radiotherapy Treatment Planning system
2568 (Alaei, 2010). If this technology is available, the patient organ doses that combine the
2569 imaging dose and the radiotherapy dose can be optimised in 3D, to create a more precise
2570 estimate of the patient's total radiation burden.

2571 (156) In summary, for most radiation oncology applications of CBCT, accurate delineation
2572 and alignment of the treatment target and critical organs should be a practitioner's primary
2573 concern. Radiation dose arising from the CBCT must be weighed within the context of
2574 therapy doses that are 1-2 orders of magnitude higher than the imaging doses. Imaging
2575 technique should be chosen to match treatment goals, such as the use of low-dose techniques
2576 for alignment of bony structures. In situations where the cumulative CBCT dose adds up to be
2577 a non-negligible fraction, it may be reflected in the overall dose schedule and subtracted from
2578 the therapeutic dose.

2579 (157) Imaging technique should be chosen to match treatment goals, such as the use of
2580 low-dose techniques for alignment of bony structures.

2581

7.3. Neurointerventions

2583

2584 (158) Intraprocedural CT capability in a C-arm, a form of CBCT, has been found to be
2585 useful in both diagnostic and therapeutic interventions. In C-arm CT, the same imaging chain
2586 that is used for fluoroscopic as well as angiographic imaging is also used for collecting the
2587 projection data needed for tomographic reconstruction.

2588 (159) CBCT is used in neurointerventions to acquire 3D angiographic images to assess
2589 potential intracranial haemorrhage, and during vertebral augmentation procedures
2590 (Psychogios et al., 2010). CBCT may also be used to guide complex, 3D positioning of coils
2591 within an aneurysm (Levitt et al., 2011). Some systems also allow over-laying of 3D images
2592 on fluoroscopic images (Racadio et al., 2007). It is even possible to create a blood-volume
2593 map with data from CT perfusion using CBCT (Fiorella et al., 2013).

2594 (160) Manufactures may provide high- and low-quality protocols for these applications.
2595 Low-quality scan protocols, which typically use a fewer number of projections, are usually
2596 sufficient for high-contrast structures such as contrast-enhanced vessels or bony anatomy.
2597 Furthermore, the position of intervention instruments can be assessed by low-dose scans. A
2598 high-quality imaging protocol is recommended for soft tissue evaluation such as assessment
2599 of intracranial parenchymal or subarachnoid haemorrhage.

2600 (161) The image quality of neurointerventional CT with respect to radiation dose using
2601 phantoms was described by Fahrig et al. (2006).

2602

2603 Table 7.2. Doses in CBCT procedures in neurointerventions.

Procedure	Reported value	Measurement technique	Reference
Head CBCT scan	Doses for brain, lens, salivary glands within scan range were between 2 and 37 mGy, effective dose was 1.2 mSv	Photodiodes in anthropomorphic phantom	Koyama et al., 2010

Neurointerventions (Soft tissue/"rule out haemorrhage")	40-48 mGy	Modified CTDI (small-volume ion chamber)	Fahrig et al., 2006
Neurointerventions (Soft tissue/"rule out haemorrhage")	75 mGy	Modified CTDI (250-mm-long ion chamber)	Kyriakou et al., 2008
Interventional head and neck surgery Soft-tissue of head and neck	10 mGy	Modified CTDI (using customised 16-cm cylindrical head phantom)	Daly et al., 2006
Neurointerventions (Angiogramms, interarterial contrast media injections)	9 mGy	Modified CTDI (250-mm-long ion chamber)	Kyriakou et al., 2008
Spine	Thoracic bone visualisation 1.8 mGy; lumbar bone visualisation 3.2 mGy; thoracic soft-tissue visualisation 4.3 mGy	Modified CTDI using CTDI (head/body) and other (abdomen/thorax) phantoms, small-volume ionisation chamber	Schafer et al., 2011
Thoracolumbar spine	Effective dose: 3.24 mSv (small patient setting), 8.09 mSv (large patient setting).	Thoracolumbar spine model, using conversion factors based on DLP	Lange et al., 2013
Neurointerventions	Brain dose: 32 mGy (high-dose CBCT)	Mathematic model of an adult standard anthropomorphic phantom	Sanchez et al., 2014

(162) In many neurointerventional scans, the radiosensitive thyroid and the eye lenses lie within the scan FOV. To minimise the dose to these organs, the user can take advantage of a feature of CBCT that is available in some MDCT scanners only as add-on feature. CBCT projections acquired over an angular span of $(180° + \varphi)$, where φ is the cone-angle of the x-ray tube, are sufficient for image reconstruction. Depending on the starting position of the $(180° + \varphi)$ rotation arc, a significant reduction in the exposure of the eyes and thyroid can be realised with "tube under" scan arcs. A shielding of the thyroid (when not in the scan FOV) provides moderate dose reduction (Daly et al., 2006).

(163) A neurointerventionalist can significantly influence the radiation dose from CBCT using the following:

1. Deciding whether or not a "high"-dose soft tissue scan is needed. This would be required to rule out intracranial haemorrhage or assess a soft-tissue structure in a diagnostic scan. For angiographic scans, for which contrast media have been injected, a "low-dose" scan that displays high-contrast structures is sufficient to image vessels. A low-dose scan is also sufficient for defining the position of high-contrast interventional materials, such as coils, clips, and Onyx (™). The choice of low vs. high dose may alter the applied dose considerably (Table 7.2.).

2623   2. Using "tube under" scans, meaning scans in which the x-ray tube is positioned on the
2624   opposite side of the body from radiosensitive organs such as the thyroid and the eyes
2625   for the majority of the time, whenever possible in practical situations. This decreases
2626   the dose to the radiosensitive organs without any appreciable consequence for the image
2627   quality or diagnostic power of the examination.

2628
2629 **7.3.1. Dose to workers from CBCT in neuroradiology procedures**
2630

2631   (164) Worker can drastically reduce their radiation exposure by maintaining sufficient
2632 distance from the x-ray source and should use shielding whenever possible. For example, the
2633 in-room unshielded effective dose from a typical intra-interventional CBCT scan (10 mGy to
2634 isocentre) is <0.005 mSv at 2 metres from the isocentre (Daly et al., 2006). Nottmeier et al.
2635 (2013) reported doses ranging between 0-70 µGy/spin and 1.8 mGy/spin in badges located at
2636 different places around the O-arm under investigation.
2637   (165) Worker should leave the room whenever permitted by the status of the patient during
2638 CBCT.

2639
2640 <div align="center">**7.4. Vascular interventions**</div>
2641

2642   (166) Vascular interventions include a range of procedures, such as angioplasty in
2643 peripheral artery disease, (fenestrated branched) endovascular aneurysm repair
2644 (EVAR/FEVAR), vessel occlusion for controlling acute bleeding, treatment of arterio-venous
2645 malformations (AVMs), and tumour embolisation, either bland (such as that in uterine fibroid
2646 embolisation), with chemotherapy (such as that in chemoembolisation of many liver tumours),
2647 or embolisation with radioactive particles (called selective internal radiotherapy treatment or
2648 SIRT). Other examples of such interventions include placement of intravascular components
2649 such as vena caval filters, transjugular intrahepatic portosystemic shunt (TIPSS), and
2650 catheter-directed thrombolysis. CBCT may be used in these procedures to acquire
2651 tomographic images of the vasculature for 3D roadmapping. CBCT is also helpful in
2652 verifying the spatial relationship of instruments and surrounding anatomy in situations where
2653 relative position or orientation cannot be resolved sufficiently using projective imaging alone.
2654 CBCT is being increasingly used for procedural planning (e.g. in trans-catheter aortic valve
2655 implantation) or image guidance and navigation [e.g. in atrial catheter ablation or TIPSS
2656 (Adamus, 2009)]. Some of the newer machines also allow acquisition of 3D vascular
2657 roadmaps that can be overlaid on fluoroscopic images. Both intra-arterial as well as
2658 intravenous contrast media injections are used. It can be expected that CBCT will play a
2659 growing role in vascular interventions.
2660   (167) The user of CBCT in vascular interventions can significantly influence the radiation
2661 dose imparted to the patient by judiciously using protocols with an adequate image quality,
2662 but lower dose, if high-contrast objects are visualised (stents, coils, guide wires or high
2663 intravascular iodine contrast), or high dose if low-contrast objects are visualised (soft tissue
2664 or low parenchymal iodine contrast).

2665
2666 Table 7.3. Patient doses in vascular CBCT interventions.

Procedure	Reported values to patient	Method	Reference
Fenestrated branched	0.27 Gy	Skin dose	Dijkstra et

endovascular aneurysm repair (FEVAR) Preoperative CBCT			al., 2011
Fenestrated branched endovascular aneurysm repair (FEVAR) Postoperative CBCT	0.552 Gy	Mean skin dose	Dijkstra et al., 2011
Catheter ablation (CBCT part)	7.9 ± 0.6 mSv	Effective dose derived from total KAP	Ejima et al., 2010
Catheter ablation (CBCT part)	5.5 ± 1.4 mSv (ICRP 60) 6.6 ± 1.8 mSv (ICRP 103)	Effective dose from simulation	Wielandts et al., 2010
Liver (in hepatic arterial embolisation therapy)	8.17 ±1.35 mSv (male) and 5.59 ±1.15 mSv (female)	Effective dose from KAP of RANDO man and woman	Tyan et al., 2013
	61.0 Gy cm^2 (male) and 52.2 Gy cm^2 (female)	KAP from 125 patients	
	11.5 ± 2.3 mSv (male) and 11.3 ± 3.0 mSv (female)	Effective dose corresponding to patients' KAP, using conversion factors based on RANDO phantoms	
Hepatic arterial embolisation therapy	75 - 175 mGy skin entry dose 16 - 52 Gy cm^2 KAP	Retrospective analysis of 126 procedures	Paul et al., 2013a Paul et al., 2013b
Abdominal CBCT scan	4-5 mSv (effective dose)	Photodiodes	Koyama et al., 2010
Abdominal CBCT	2.1-4.2 mSv (effective dose)	"Small" anthropomorphic phantom and Monte-Carlo simulations	Suzuki et al., 2011
Hepatic artery embolisation	238 mGy (skin dose)	Skin entry dose readout from examination protocol	Schulz et al., 2012

7.4.1. Dose to worker in vascular interventions.

(168) Paul et al. (2013b) found that the dose to the hands and the left knee of the interventionalist was higher than those of the assistant physician when using volume imaging. Mean doses received by the interventionalist ranged from 0.01 mGy to the shielded thyroid, chest and gonads, to 0.37 mGy to the left finger. The corresponding dose range for the assistant physician was from 0.01 mGy to the shielded thyroid, chest and gonads, to 0.08

2675 mGy to the left and right eyes. The mean eye doses for the interventionalist were 0.11 mGy.
2676 Doses associated with the use of CBCT were higher as compared to catheter angiography and
2677 DSA. In guided needle interventions, operator hand doses in free-hand procedures ranged
2678 from 20–603 μSv. Laser guidance alone or in combination with needle holders resulted in a
2679 reduction of the hand dose to <36 μSv (5–82 μSv) per procedure (Kroes et al., 2013).

2680 (169) Worker should leave the room whenever permitted by the clinical situation during a
2681 CBCT scan. For injecting contrast media, an automatic injector should be used whenever
2682 possible. Personnel who remain in the procedure room during the CBCT exposure should be
2683 protected by fixed or mobile shields.
2684
2685 Table 7.4. Woker doses in vascular CBCT interventions.

Procedure	Reported Value to worker	Method	Reference
Abdominal CBCT	Eye level: 8 seconds/rotation: 28.0 μSv, 20 seconds/rotation: 79.3 μSv, 5 seconds/2 rotations: 32.5 μSv, large FOV 37.6 μSv	Digital dose rate meter at different positions in the room	Schulz et al., 2012
Hepatic angiography	Eye level: 28-79 μSv per procedure	Digital dose rate meter at different positions in the room	Schulz et al., 2012

2686
2687 **7.5. Non-vascular interventions**
2688

2689 (170) Non-vascular interventions include procedures such as vertebroplasty (treatment of
2690 vertebral fractures, osteoporosis or metastases), drainages of abscesses or fluid collections,
2691 image-guided biopsies, percutaneous transhepatic cholangiography drainage (PTCD), and
2692 tumour ablation (e.g. liver tumour microwave ablation) (Wallace et al., 2008). Those
2693 procedures are currently performed either under fluoroscopic guidance or MDCT-guidance,
2694 with C-arm CBCT becoming increasingly popular as it combines advantages of both (Orth et
2695 al., 2008). Modern C-arm systems allow the planning of percutaneous instrument insertion
2696 via a pre-procedural CBCT with fluoroscopy as the main modality for intra-procedural
2697 instrument guidance. Repeated CBCT may be used for intra-procedural quality control;
2698 however, the user should minimise the number of CBCT scans acquired during a given
2699 procedure.

2700 (171) The user of CBCT in non-vascular interventions can significantly influence the
2701 radiation dose that is applied to the patient by:

2702 • Appropriately choosing between a "high-dose" vs. "low-dose" scan; and
2703 • Judiciously using the CBCT mode, relying on the fluoroscopy mode as far as possible.

2704 (172) Table 7.5. provides an overview of patient doses in non-vascular interventions.
2705 Doses vary considerably depending on the diagnostic application and corresponding exposure
2706 settings. Effective doses measured in phantoms were a few mSv for each study. Various other
2707 dose quantities are also included. Reported CTDI values were generally a few mGy, but some
2708 values >20mGy have been measured. At the skin and eye level, doses up to a few hundred
2709 mGy were found.

2710
2711 Table 7.5. Patient doses in non-vascular CBCT interventions.

Procedure	Reported values to phantom representing patient	Method	Reference

Lumbar spine (bone protocol)	3.70 mGy	Modified CTDI*	Schafer et al., 2011
Thoracic spine (bone protocol)	1.91 mGy	Modified CTDI*	Schafer et al., 2011
Lumbar spine low resolution (soft tissue protocol)	6.01 mGy	Modified CTDI*	Schafer et al., 2011
Lumbar spine high resolution (soft tissue protocol)	12.50 mGy	Modified CTDI*	Schafer et al., 2011
Thoracic spine (soft tissue protocol)	4.61 mGy	Modified CTDI*	Schafer et al., 2011
CBCT-guided vertebroplasty of the thoracic spine	11.5 mGy (total procedure dose)	Modified CTDI*	Schafer et al., 2011
CBCT-guided vertebroplasty of the lumbar spine	23.2 mGy	Modified CTDI*	Schafer et al., 2011
Renal Biopsy	44.0 Gy cm^2	Mean KAP	Braak et al., 2012
Biliary tube placement (PTCD)	413 mGy	Skin entrance dose	Schulz et al., 2012
"Biliary protocol"	4.2-8.4 mSv (effective dose)	Female anthropomorphic phantom with MOSFET detectors	Kim et al., 2011
Phantom study	Head: 1.18 mSv Chest: 7.32 mSv Abdomen: 7.48 mSv	TLDs in Alderson phantom	Bai et al., 2011
Head and abdominal imaging comparison of CBCT to MDCT	Head protocol: 4.4-5.4 mSv (Eye doses: 44.6-173.6 mGy) Abdominal Protocols: 15.0-37.0 mSv	Effective dose estimates measured with TLDs in a dosimetric phantom	Kwok et al., 2013

2712 *Using CTDI (head/body) and oblate (abdomen/thorax) phantoms, measuring at central and
2713 four peripheral points with a small-volume ionisation chamber.
2714
2715 **7.5.1. Dose to worker in non-vascular interventions**
2716
2717 (173) In certain procedures, some dose to the interventionalist cannot be avoided. For
2718 example, percutaneous transhepatic cholangiography (PTC), cholangial drainage (PTCD), or
2719 other biliary drainage (PTBD) procedures often require that one or both hands/fingers are

2720 very close to the radiation field. For a short time, these procedures may even require that
2721 these organs be in the radiation field, especially in punctures of the left lobe of the liver. The
2722 practitioner should be cognisant of these small but potentially repeated exposures. In a long
2723 procedure, the dose to the fingers may exceed a few mSv. Protective gloves reduce the
2724 exposure of hands or fingers but increase the dose of worker and patient if the hands with
2725 gloves are placed in the primary beam. Auxiliary instrumentation for handling needles and
2726 probes in the radiation field should be used whenever possible. Examples of doses to worker
2727 from interventional procedures are given in section 7.4.1; radiation doses in vascular and non-
2728 vascular interventions are similar.
2729
2730 **7.6. Orthopaedics/Surgery**
2731
2732 (174) In orthopaedics or trauma surgery, CBCT is used mainly to assess the position of
2733 fractures and implants with respect to the bony anatomy, especially in situations where
2734 fluoroscopy alone is insufficient to disambiguate the position of an implant with respect to the
2735 bony anatomy. For example, with fluoroscopy alone, the critical relationship of a screw with
2736 respect to an articular surface may sometimes remain unclear. CBCT may be a big help in
2737 clarifying this relationship. CBCT is also very helpful in spine surgery where interventions
2738 are being performed in close proximity to critical structures such as spinal nerves. CBCT
2739 datasets are also used to confirm the position of implants inter-procedurally or to acquire
2740 datasets for intraoperative navigation. Dedicated extremity CBCT systems are based on the
2741 same principle as other CBCTs used in interventional radiology or elsewhere, with C-arm
2742 being the most popular platform. Another system called the O-arm is becoming increasingly
2743 popular for extremity and spinal fixation procedures. An O-arm system combines the
2744 advantages of a CT-gantry based design with the flexibility of a C-arm based design. It is
2745 essentially a C-arm system with a telescopic gantry that extends out to complete the ring and
2746 become an O-arm for CT operation. As such, the gantry can function as a standard C-arm, or
2747 one can complete the O-ring, and turn the system into a CT-like gantry where the FPD and
2748 the x-ray tube freely rotate. Usually, CBCT scanning is performed intra-operatively in a prone
2749 or supine position. Standing position for imaging of knee weight-bearing position, or while
2750 the patient is sitting with the upper or lower extremities extended (Zbijewski et al., 2011),
2751 have been described (Tuominen et al., 2013).
2752
2753 Table 7.6. Patient doses in orthopaedics/surgery CBCT interventions.

Procedure	Reported values to patient	Method	Reference
Extremity scan	0.064-0.15 mSv	Modified CTDI approach	Zbijewski et al., 2011
CBCT wrist arthrography	2.1 mGy	Modified CTDI	Ramdhian-Wihlm et al., 2012
Evaluation of finger fractures	0.8 mSv	TLDs absorbed tissue dose	Faccioli et al., 2010
volumetric scan of wrist joint and the distal radius	0.11 mSv	Modified CTDI	Reichardt et al., 2008

| Spine | 1.8 mGy (thoracic "bony" spine)
3.2 mGy (lumbar "bony" spine)
10.6 mGy (soft-tissue, high-spatial resolution)
5.1 mGy (soft-tissue, low-spatial resolution) | QRM phantoms, modified CTDI approach, ionisation chambers | Schafer et al., 2011 |
| Spine, Vertebroplasty | Thoracic 11.5 mGy
Lumbar 23.2 mGy | Cumulative dose of QRM phantoms, ionisation chambers | Schafer et al., 2011 |

7.7. Urology

(175) CBCT on a C-arm also enables cross-sectional imaging to be performed in a urological operating room. Apart from the standard pulsed fluoroscopy, 3D reconstruction can be performed intra-operatively during urologic procedures. Different operating modes are available. A low-dose protocol may be appropriate when imaging high-contrast structures. For example, when imaging calcified stones or other calcifications during percutaneous nephrolithotomy, a low-dose protocol should be employed because kidney stones should be visible despite high noise in the images obtained. The same reasoning holds true for CBCT imaging of retrograde flow of contrast in the urinary tract and collecting system (Michel et al., 2014; Roy et al., 2012).

(176) The user should use low-dose protocols that are sufficient to detect kidney stones, pelvic calcifications, metallic instrumentation, and contrast media filled efferent urinary tract.

7.8. ENT and head diagnostics or surgery

(177) Similar to other applications in the head and neck area, applications of CBCT in ENT take advantage of the fact that this region includes structures, such as the paranasal sinuses, the temporal bone and the skull base that have high intrinsic contrast, being composed primarily of bone, air, and soft tissue. Therefore, relatively high noise in the images can be tolerated without compromising the diagnostic utility of the CBCT scans. The high-resolution of CBCT systems is ideally suited for the small structures of the skull base and middle ear. In addition, only a relatively small scan FOV is required to cover the necessary anatomy. In ENT scans, the position of the scan-arc is a significant factor that influences radiation exposure of sensitive organs such as the eye lens and thyroid (Daly et al., 2006). Other applications of CBCT in ENT are described in (Hodez et al., 2011; Miracle and Mukherji, 2009). For most diagnostic ENT procedures such as imaging of the temporal bone and paranasal sinuses, dedicated scanners with the patient in a sitting position are used. Besides low-dose and patient comfort, high spatial resolution is another major advantage of these scanners. As a result, these scanners are increasingly being used for surgical planning of temporal bone interventions such as cochlear implantation. There has been a rapid adoption of this technology in routine clinical practice, a trend that is likely to accelerate in future.

Table 7.7. Patient CBCT doses in ENT and head surgery.

Procedure	Reported values to patient	Methods	Reference

"Head scan mode" – soft-tissue mode	10 mGy	Modified CTDI (custom 16-cm cylindrical head phantom)	Daly et al., 2006
Sinus imaging (bone mode)	3 mGy and above	Modified CTDI (custom 16-cm cylindrical head phantom)	Daly et al., 2006
Endoscopic sinus surgery	10.7 mGy ±0.6 mGy	CT head phantom was used along with a ion chamber	Manarey et al., 2007
Head CBCT protocols dose compared to MDCT	MDCT head protocol: • CBCT: 4.4-5.4 mSv • MDCT: 4.3 mSv	Effective dose estimates measured with TLDs in a dosimetric phantom	Kwok et al., 2013

2789
2790   **7.9. Dental (oral and maxillofacial)**
2791

2792   (178) CBCT has been used in oral and maxillofacial imaging for several years, and its use
2793 is increasing. It is primarily used to acquire images of the teeth and periodontium, their
2794 placement within the alveolus of the mandible and maxilla, and their relationship with the
2795 adjacent nerves and other structures. The high spatial resolution of CBCT is ideally suited for
2796 these high-contrast structures and generally provides excellent image quality in this field. The
2797 images are used for diagnostic purposes, pre-operative planning, and image-guidance during
2798 navigated surgery in this region. Pathological changes such as fractures, periapical abscesses,
2799 caries or periodontal disease affect high-contrast structures and can therefore be imaged
2800 precisely using CBCT. The FOV is usually large enough to cover the maxillofacial region
2801 with one orbit around the patient. In addition, dedicated small volumes (e.g. 4 × 4 cm) allow
2802 for high-resolution imaging of a small region of interest, such as a single tooth root, at a very
2803 low radiation dose. Earlier scanners employed image intensifiers, but in the current systems,
2804 FPDs are being used almost exclusively. Most systems are seat-scanners consisting of a small
2805 C-arm that rotates in a horizontal plane along a vertical axis with the patient sitting upright.
2806 Applications of dental CBCT are described in De Vos et al. (2009).
2807   (179) Due to the wide dose range found in dental CBCT and the variety of diagnostic
2808 needs in dental radiology, proper application of this technique among alternative 2D and 3D
2809 dental imaging modalities has been of great concern since its introduction in dentistry in 1998.
2810 Owing to its relatively low radiation dose and high spatial resolution compared to MDCT,
2811 dental CBCT is considered as a suitable substitute for MDCT for several applications (e.g.
2812 implant planning). However, its application as a complement or substitute for 2D imaging
2813 modalities (e.g. panoramic or cephalometric radiographic) increases the population dose. In
2814 many cases such as the detection of root pathology, CBCT has superior diagnostic efficacy
2815 compared with 2D radiographs; but for other applications, such as the pre-operative
2816 evaluation of third molars, 2D radiographs often suffice. Detailed evidence-based guidelines
2817 have been determined during the SEDENTEXCT project and have been published in
2818 Publication 172 of the European Commission (EC, 2012). The guidelines encompass a
2819 variety of topics, covering justification, optimisation, training and QA aspects. Twenty "Basic
2820 Principles" were defined based on a thorough literature review in combination with the
2821 experimental work performed in SEDENTEXCT on radiation dose, diagnostic use and other
2822 CBCT-related topics.

2823   (180)   Several basic principles relate to justification, as the excessive use of CBCT in
2824 dentistry would increase the population dose. The use of CBCT in dentistry can only be
2825 considered as justified, if a patient history and clinical information are available, if it is
2826 expected to add new information, and if 2D radiographs do not (or are not expected to)
2827 answer the diagnostic question. Repeated CBCT examinations should be avoided unless each
2828 examination can be individually justified. In addition, CBCT should not be used if soft tissue
2829 assessment is required, since only MDCT or MRI provides the contrast resolution required
2830 for soft tissue imaging.

2831   (181)   An important optimisation principle in dental CBCT relates to the choice of the
2832 appropriate volume size for each examination. In many cases, the region of interest is known
2833 exactly before scanning; in other cases, the required volume is revealed after acquisition of a
2834 frontal and lateral scout image. The smallest available volume size should always be chosen,
2835 as this could greatly reduce patient dose. The choice between high- and low-dose settings
2836 should be made according to the optimisation principle, ensuring adequate image quality for
2837 diagnosis at the lowest achievable dose.

2838   (182)   Since CBCT images often contain structures that are not part of the diagnostic region
2839 of interest (although this should be limited as much as possible through FOV reduction), the
2840 EC guidelines also state that the entire image should be examined and reported, not just the
2841 region of interest. Depending on the scanning region, the involvement of an oral or medical
2842 radiologist can be warranted.

2843   (183)   Table 7.8. provides an overview of the effective dose range in dental CBCT,
2844 measured using anthropomorphic phantoms. Although accuracy and intercomparability of
2845 several dosimetric studies are limited due to the varying measurement methodology (e.g.
2846 TLD placement), the table shows that patient doses vary considerably, which is a direct result
2847 of the wide variation of exposure parameters being applied. Volume sizes range between a
2848 few cm^3, sufficient for scanning of a single tooth area, and a few thousand cm^3, covering most
2849 of the head. In addition, there is no standardisation regarding the kVp used in dental CBCT,
2850 with values ranging between 70 and 120 kV. Clinically applied mAs values range more than
2851 20-fold but are mostly found between 25 and 150 mAs.

2852

2853 Table 7.8. Overview of radiation doses in dental CBCT (Source: EC Radiation Protection Publication
2854 172, 2012).

Dental CBCT unit type	Effective dose (µSv)
Dento-alveolar	11-674 (median: 61)
Craniofacial	30-1073 (median: 87)

2855

2856   (184)   The application of dental CBCT for paediatric patients is of particular concern due to
2857 their higher radiosensitivity. Similar to its adult applications, paediatric use of CBCT could
2858 lead to considerable dose reduction when used as a replacement to MDCT (e.g. cleft palate),
2859 providing that FOV limitation is applied and that exposure factors are optimised. However, its
2860 use as a complement to or replacement for 2D radiography could lead to patient doses which
2861 are disproportionate to the diagnostic benefit, especially when large-volume coverage is
2862 required (e.g. orthodontic planning). For most paediatric applications, more evidence
2863 regarding diagnostic efficacy of CBCT is needed before widespread application can be
2864 considered. Table 7.9. contains effective dose measurements for 10 year-old and adolescent
2865 anthropomorphic phantoms. Due to the larger relative coverage of the child's head, effective

2866 doses are higher compared with adults if exposure factors are not adapted. For some CBCT
2867 models, pre-set "child dose" exposure parameters are available, typically corresponding to a
2868 reduction in mAs. For other models, exposure factors can be modified by the operator. AEC
2869 is largely absent in dental CBCT, with one manufacturer having applied it for several years.
2870

2871 Table 7.9. Overview of radiation doses in dental CBCT for different patient ages (Source: EC
2872 Radiation Protection Publication 172, 2012).

Age	Dental CBCT unit type	Effective dose (μSv)
10 year-old phantom	Dento-alveolar	16-214 (median: 43)
10 year-old phantom	Craniofacial	114-282 (median: 186)
Adolescent phantom	Dento-alveolar	18-70 (median: 32)
Adolescent phantom	Craniofacial	81-216 (median: 135)

2873
2874 (185) Corresponding with the wide range in effective dose, absorbed doses of 0.03-10.0
2875 mGy have been reported for the thyroid gland, 0.02-9.3 mGy for the brain and 0.03-16.7 mGy
2876 for the eye lens (Hirsch et al., 2008; Ludlow and Ivanovic, 2008; Ludlow et al., 2006;
2877 Pauwels et al., 2012). Various dose indices have been measured in dental CBCT as well. A
2878 2009 report by the United Kingdom (UK) Health Protection Agency (HPA) measured KAP
2879 for 41 dental CBCTs and normalised the results to a 4 × 4 cm field size, with values ranging
2880 between <100 and >2300 mGy.cm^2 (HPA, 2010).
2881 (186) Exposure of the worker is reported to be in the range of 2 to 40 μGy per scan at 1
2882 metre. For comparison, intraoral and panoramic radiography scatter doses are less than 1 μGy
2883 per exposure at 1 metre (EC, 2012). The EC guidelines on dental CBCT state that "for worker
2884 protection from CBCT equipment, the guidelines detailed in Section 6 of the European
2885 Commission document 'Radiation Protection 136. European Guidelines on Radiation
2886 Protection in Dental Radiology' should be followed".
2887

2888 **7.10. Breast**
2889

2890 (187) Mammography has been the standard imaging method for breast cancer screening
2891 for three decades. While digital mammography has replaced screen-film mammography in
2892 many locations, the projection-imaging nature of mammography did not change with the
2893 introduction of digital mammography; digital mammography still requires compression of the
2894 breast in order to acquire a 2D projection image of the 3D breast. Digital mammography was
2895 proven to be slightly more effective in detection of small lesions in women under 50 years old
2896 with radiographically dense breasts (Pisano et al., 2005). Digital mammography has also been
2897 shown to reduce breast dose in comparison to screen-film radiography. In a 2010 study, mean
2898 glandular dose per view averaged 2.37 mGy for screen-film mammography while it was 22%
2899 lower (1.86 mGy per view) for digital mammography (Hendrick et al., 2010). With digital
2900 mammography, contrast can be restored (within limits) using digital enhancement techniques.
2901 Therefore, a harder x-ray spectrum can be used with digital mammography compared to
2902 screen-film mammography, and this is the primary reason that some dose reduction is
2903 possible. The harder x-ray spectrum is achieved through the use of different anode/filter
2904 combinations (e.g.tungsten/rhodium instead of molybdenum/molybdenum) and higher
2905 average tube potentials.

(188) 2D mammography suffers from the superposition of structures that may falsely appear normal or abnormal, and this anatomical noise created by the normal parenchyma of the breast confounds the cancer detection task. 3D approaches relying on the principles of CT may improve breast cancer detection, especially in the dense breast. Two approaches for "3D" imaging of the breast have been proposed: digital breast tomosynthesis; and bCT. Breast tomosynthesis is performed using multiple (e.g. 15–30) low-dose digital 2D projection images, acquired on a modified full-field digital mammographic system which allows limited angular movement of the x-ray tube around the breast during acquisition (Poplack et al., 2007; Niklason et al., 1997). Tomosynthesis is the name given to this acquisition strategy, which is formally considered to be limited-angle tomography.

(189) Patient dose in one breast tomosynthesis acquisition, comprising 11 low-dose projections over 28 degrees angular movement, is approximately 4 mSv for a breast of average thickness. This is about twice the dose used for digital mammography (Poplack et al., 2007). More recently, doses from breast tomosynthesis were estimated to be between 1.66 and 1.90 mGy for a standard breast, based on manufacturer's data in the absence of a standard protocol (Michell et al., 2012). More recent tomosynthesis systems use a number of x-ray projections whose cumulative dose to the breast is comparable to conventional single-view digital mammography.

(190) bCT is currently undergoing evaluation before it can be introduced into clinical practice. This technology has been developed to address the shortcomings of conventional mammography such as contrast resolution and the problems occurring from overlap of structures in 2D images (O'Connel et al., 2010). Most bCT systems make use of FPDs, and therefore are CBCT systems; however, helical CT systems for dedicated breast imaging (Kalender et al., 2012) are also being designed.

(191) In the early days of bCT, there was no established method for estimating the mean glandular dose to the breast in the pendant geometry used for this modality. Therefore, methods for computing the dose to the breast needed to be developed. Monte Carlo techniques were used to develop comprehensive tables of so-called DgN_{CT} values, which are appropriate for 360° scanning of the pendant breast (Boone et al., 2004; Boone et al., 2005).

(192) Cone beam-based bCT systems use FPDs that acquire 2D projections which completely encircle the breast. Typically, a complete breast scan (of a single breast) requires from 10 to 17 seconds, and about 300-500 projections are acquired within this time (O'Connel et al., 2010; Packard et al., 2012). These systems are designed to be low dose, and the mean glandular dose can be as low as that of two view mammography for each woman. Obviously, radiation dose depends on breast size and composition. Therefore, smaller doses will occur in smaller breasts, and larger breasts will receive higher doses. Reported mean glandular dose values range between 4-12.8 mGy (O'Connell et al., 2010) and 2.5-10.3 mGy (Lindfors et al., 2008). Average doses from conventional mammography documented in the above mentioned study by O'Connell et al. (2010) were in the range of 2.2-15 mGy.

(193) Currently, bCT technology has some limitations regarding the detection of microcalcifications as well as coverage of the axillary region, both of which are performed better with conventional mammography (Lindfors et al., 2010; O'Connell et al., 2010). Higher resolution detector systems will likely improve spatial resolution of bCT and consequently improve microcalcification detection performance as well (Kalender et al., 2012).

(194) Worker dose considerations for bCT are minimal since the worker does not need to be near the patient during image acquisition, as with most CT settings. Of course, proper shielding of the bCT room is considered to be essential. One issue in regards to shielding will emerge if bCT scanners become more commonplace in the clinical imaging environment.

These systems make use of higher energy x-ray spectra than mammography systems, and therefore, it is likely that additional room shielding will be required if a bCT system is installed in a mammography room. Please see Chapter 3 for more details on room shielding.

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CRP</image> **DRAFT REPORT FOR CONSULTATION:** DO NOT REFERENCE

3173
3174 **8. TRAINING CONSIDERATIONS FOR CBCT**
3175
3176 • **The recommendations provided by the Commission on education and training in its**
3177 **_Publication 113_ are applicable here for CBCT.**

3178 • **The level of training in radiological protection should be commensurate with the level**
3179 **of expected radiation exposure (ICRP, 2009).**

3180 • **All personnel intending to use CBCT for diagnostic purpose should be trained in the**
3181 **same manner as for diagnostic CT and for interventional CBCT same as**
3182 **interventional procedures using interventional CT.**

3183
3184 **8.1. Introduction**
3185

3186 (195) The ICRP, in its _Publication 113_ (ICRP, 2009), provides substantial information and
3187 guidance on training of health professionals in radiological protection for diagnostic and
3188 interventional procedures. Much of the information provided in this section is derived from
3189 this publication.

3190 (196) The ICRP states that a training programme in radiological protection for healthcare
3191 professionals has to be oriented towards the type of practice in which the target audience is
3192 involved (ICRP 2009; ICRP 2010).

3193 (197) The main purpose of training is to make a qualitative change in practice that helps
3194 operators use radiological protection principles, tools, and techniques to reduce their own
3195 exposure without cutting down on work, and to reduce patient exposure without
3196 compromising on image quality or intended clinical purpose. The focus has to remain on
3197 achievement of skills. Unfortunately, in many situations, training takes the form of complying
3198 with requirements of number of hours. While the number of hours of training provides an
3199 important yardstick, it is also essential to require trainees to learn skills to reduce
3200 occupational and patient exposure. In large parts of the world, clinical professionals engaged
3201 in the use of radiation outside imaging departments have either no training or inadequate
3202 training. The Commission has recommended that the levels of education and training should
3203 be commensurate with the level of radiation use and expected radiation exposure (ICRP,
3204 2009). As the use of CBCT outside imaging departments increases, the need for education
3205 and training of personnel also increases. Professionals who are directly involved in operation
3206 of CBCT for diagnosis or intervention and interpreting CBCT studies should receive
3207 education and training in radiological protection at the start of their career, and refreshment
3208 and professional development training should continue throughout their professional life.
3209 Continuing education should include specific training on relevant radiological protection
3210 tools and procedures as new equipment or techniques are introduced.

3211 (198) Legislation in most countries requires that individuals who take responsibility for
3212 medical exposures must be properly trained in radiological protection.

3213 (199) Training activities in radiological protection should be followed by an evaluation of
3214 the knowledge acquired from the training programme (a formal examination system).

3215 (200) Personnel who have completed training should be able to demonstrate that they
3216 possess the knowledge specified by the curriculum by passing an appropriate certifying
3217 examination.

3218 (201) Nurses and other healthcare professionals who assist during CBCT procedures
3219 should be familiar with radiation risks and radiological protection principles in order to
3220 minimise their own exposure and that of others.

3221   (202)   Medical physicists should become familiar with the clinical aspects of the specific
3222 procedures performed at their local facility.

3223   (203)   The issue of delivery of training and assessment of competency has been dealt with
3224 in *Publication 113* (ICRP, 2009).

3225

8.2. Curriculum

3227

3228   (204)   It is anticipated that a large fraction of professionals involved in CBCT will be those
3229 who have prior education in medical radiation physics and radiological protection. Thus,
3230 simple orientation training may suffice in such cases. All personnel intending to use CBCT
3231 for diagnostic purpose should be trained in the same manner as for diagnostic CT and for
3232 interventional CBCT the same as interventional MDCT keeping the level of dose and usage
3233 in view as specified earlier.

3234   (205)   It has been observed that most organisations follow the relatively easy route of
3235 requiring a certain number of hours of education and training. The Commission gives some
3236 recommendations on the number of hours required, but this should act as a guideline and not
3237 be applied rigidly (ICRP, 2009). Providing guidance in terms of the number of hours has
3238 advantages in terms of implementation of training and monitoring the training activity, but is
3239 only a guide.

3240   (206)   Many programmes fail with regard to assessment of whether the objectives have
3241 been achieved. Others have pre- and post-training evaluations to assess the knowledge gained,
3242 but few training programmes assess the acquisition of practical skills. It would be more
3243 appropriate to encourage development of questionnaires and examination systems that assess
3244 knowledge and skills, rather than prescribing the number of hours of training. The extent of
3245 training depends upon the level of radiation employed in the work, and the likelihood of
3246 overexposure to the patient or workers.

3247

8.3. Who should be the trainer?

3249

3250   (207)   In view of the importance of this issue, most of the text from *Publications 113*
3251 and/or *117* is reproduced here. The foremost point in any successful training is that the trainer
3252 should have a clear perception about the practicalities of the work that the training has to
3253 cover. The primary trainer should normally be an expert in radiological protection (normally
3254 a medical physicist) and should have knowledge about clinical practice involving the use of
3255 radiation. That is, the trainer should know about the nature of radiation, the way in which it is
3256 measured, how it interacts with the tissues, what type of effects it can lead to, principles and
3257 philosophies of radiological protection, and international and national guidelines. As
3258 radiological protection is covered by legislation in almost all countries of the world,
3259 awareness of national laws and the responsibilities of individuals and organisations are
3260 essential (ICRP, 2009).

3261   (208)   Training should deal with what people can practice in their day-to-day work. Instead,
3262 many trainers in radiological protection cannot resist the temptation to talk about basic topics
3263 such as definition of radiation units, interaction of radiation with matter, and even in-depth
3264 information on structure of the atom and atomic radiation in more detail than is appropriate
3265 for the clinical audience and for the practical purposes of radiological protection training.
3266 Such topics, while being essential in basic educational programmes, should only be dealt with
3267 to a level such that they make sense in the context of radiological protection training. A
3268 successful trainer should not be too focussed on definitions which are purely academic, but
3269 should be guided by the utility of the information to the audience. The same applies to

3270 regulatory requirements. The trainer should speak the language of users to convey the
3271 necessary information without compromising on the science and regulatory requirements.
3272 Health professionals who use radiation in day-to-day work in hospitals and deliver the
3273 radiation dose to patients know about the practical problems in dealing with patients who may
3274 be very sick. They understand problems with the radiation equipment they deal with, the time
3275 constraints for dealing with large numbers of patients, and the lack of radiation measuring and
3276 radiological protection tools. It is recommended that training also includes lectures from
3277 practising clinicians and imaging specialists, who can focus on good and bad radiological
3278 protection practices. It may be useful for the radiological protection trainer to be available
3279 during such lectures to comment and discuss any issues raised.
3280
3281 <div align="center">**8.4. References**</div>
3282
3283 ICRP, 2009. Education and training in radiological protection for diagnostic and
3284   interventional procedures. ICRP Publication 113. Ann. ICRP 39(5).
3285 ICRP, 2010. Radiologicalprotection in fluoroscopically guided procedures outside the
3286   imaging   department.   ICRP   Publication   117.   Ann.   ICRP   40(6).

3287
3288   **9. QUALITY ASSURANCE PROGRAMMES**
3289

3290 • **QA Programmes for CBCT should follow guidelines outlined by international**
3291   **standards and professional societies.**

3292 • **DRLs are not yet established for most CBCT applications. In the absence of**
3293   **international or national DRLs, local DRLs should be established to inform local**
3294   **policy.**

3295
3296   **9.1. Introduction**
3297

3298   (209) The purpose of a QA programme is to ensure consistent and adequate image quality
3299 while minimising the radiation dose to the patient, and maintaining performance and safety of
3300 the equipment in conformance with specifications. In the context of this report, the QA
3301 programme consists of the acceptance and commissioning of CBCT equipment, as well as
3302 periodic test and maintenance of equipment performance, patient imaging protocols, worker
3303 and patient dose, worker training, and adherence to policies and procedures.

3304
3305   **9.2. Quality control of CBCT equipment**
3306

3307   (210) Quality control begins when the equipment is installed and continues throughout its
3308 lifetime. The acceptance test, commissioning, and status testing of equipment should ensure
3309 that the system is operational according to the manufacturer's specifications, which are based
3310 on national or international standards. At the time of acceptance, baseline measurements of
3311 image quality and dosimetry should be taken along with parameters that affect these factors.
3312 These measurements will be used as a reference for comparison with later measurements, and
3313 can indicate if the system performance has degraded and needs corrective action.
3314   (211) Equipment tests fall into six categories: safety system, x-ray generator performance,
3315 image quality, geometry, display, and dosimetry. Safety system tests are used to ensure the
3316 proper operation of warning lights, door and collision interlocks, portable shielding, and the
3317 emergency-off system. x-ray generator tests can ensure that the x-ray system operates
3318 properly, including the accurate production of kV, mA, exposure time, and linearity. Image
3319 quality tests, such as those that measure noise, uniformity, contrast, and resolution, can ensure
3320 that acquired images are suitable for clinical use. The frequency of these quantitative tests
3321 should be established to remediate image quality degradation (IEC 61223-2-6, 2006). In
3322 addition to quantitative testing, images should be visually inspected to identify image
3323 artefacts. Geometry tests are used to ensure proper system alignment and scaling. In
3324 radiotherapy applications, a daily test of the CBCT image isocentre geometry ensures that
3325 images are aligned with the treatment machine. However, dental and interventional
3326 applications may not require alignment with an external coordinate system, and therefore,
3327 need only test image scaling. Display testing will ensure that image presentation is consistent
3328 and faithful to avoid loss of information during interpretation. Finally, dosimetry tests are
3329 used to assess the dose to a phantom, using standard measurement protocols appropriate for
3330 CBCT, such as those described earlier in this document. The equipment and methods needed
3331 to perform other tests are described elsewhere (IPEM Report 91, 2005).
3332   (212) The schedule and scope of routine testing of CBCT equipment depend to some
3333 degree on the clinical application. Inspection schedules recommended by six different
3334 organisations (three for dental applications, and three for radiotherapy applications) are

3335 shown below. The schedules are largely in agreement, but some special considerations are
3336 worth noting. For CBCT equipment with an exposed moving gantry that might collide with
3337 patients or worker, a daily safety system check is recommended. If the CBCT image
3338 coordinates are used to control a radiotherapy accelerator or surgical equipment, a daily check
3339 of coordinate system integrity is recommended. If accurate density information (such as HU
3340 numbers) is used for diagnosis or planning, these values should be tested at least monthly.
3341 Users should therefore consider these general guidelines to inform a risk-based QA program
3342 based on their clinical aims.
3343
3344 Table 9.1. Proposed QA test and corresponding periodicity as recommended by international, national
3345 and professional societies.

QA Test	Daily	Monthly	Periodic	Annual
Safety systems: collision, warning lights and interlocks	142, 179, IAC			
Image quality: Uniformity		EC, 142, 179, HPA	179, IAC	
Image quality: Image density	IAC	EC, 142, 179, HPA		
Image quality: Noise	IAC	EC, 142, 179, HPA	179	
Image quality: Low contrast detail		142, 179	179, IAC	EC
Image quality: High contrast resolution		142, 179	179, IAC	EC, HPA
Image quality: Assess image artefacts	IAC	EC		
Geometry: isocentre coincidence	142, 147, ACR			
Geometry: scaling and slice thickness		142, 179	179	EC, HPA, IAC
Data storage and transfer			ACR, IAC	
Image registration software			ACR	
Image display		EC	HPA	IAC
x-ray quality, linearity, and field size				EC, 179, HPA, IAC
Dose measurements				EC, 142, 179, HPA, IAC

3346
3347 142: AAPM report 142: Klein et al., 2009.179: AAPM report 179, 2012.ACR: ACR,
3348 2009.HPA: HPA, 2010. IAC: IAC, 2012. EC: EC, 2012.
3349
3350 **9.3. Patient dose reporting**
3351

3352   (213) The need for dose reporting in CBCT follows from the principles of optimisation of
3353   radiation protection. Radiation dose to the patient cannot be optimised to as low as reasonably
3354   achievable without accurate tracking of dose. The most straightforward method for achieving
3355   dose tracking is through the electronic display of dose on the imaging console (ICRP, 2004),
3356   and recording of delivered dose into the patient record as a DICOM-structured dose report
3357   (IEC 60601-2-44, 2012).

3358   (214) In MDCT systems, it is now standard to display estimates of delivered dose directly
3359   on the console numerically as $CTDI_{vol}$ and DLP. These estimates represent the dose to a
3360   phantom, not the dose to a patient. Methods should be developed for estimating doses to
3361   patients based on patient size and the scanning parameters used for individual patients. A
3362   medical physicist, as part of the QA programme, should verify the accuracy of these numbers
3363   at least annually, or whenever equipment is repaired in a manner that can affect dose. For
3364   CBCT systems, the system for dose reporting is not yet standardised. The UK HPA (2010)
3365   and EC (2012) recommend that the dose estimate be displayed as KAP in dental CBCT
3366   systems. The QA program should be prepared to verify dose estimates as they are reported by
3367   each device, whether it be KAP or CTDI and DLP.

3368   (215) Electronic transfer of patient dose to an electronic medical record greatly facilitates
3369   the tracking of annual and lifetime radiation dose to a patient over multiple procedures.
3370   MDCT systems implement this idea using the DICOM-structured dose report, which usually
3371   expresses dose in terms of $CTDI_{vol}$ and DLP. Electronic transmission of $CTDI_{vol}$ and DLP to
3372   PACS is now required by California State law in the United States (California Senate Bill
3373   SB1237, 2010), and has been proposed by the EC (European Commission, 2011). Electronic
3374   reporting further supports initiatives to compare recorded doses with DRLs, a concept
3375   recommended by ICRP for optimisation (ICRP, 2007). Dose registries are another potential
3376   tool for facilitating evaluation of patient dose.

9.4. Diagnostic reference levels

3380   (216) DRLs have been established through government and professional organisations to
3381   guide users in optimising procedure performance for both image quality and radiation
3382   reduction. While these efforts have matured for MDCT imaging, little progress has been
3383   made toward setting DRLs for CBCT. SEDENTEXCT (EC, 2012) recommends a single
3384   reference level of 250 mGy.cm^2 for the placement of an upper first molar implant in adults.
3385   For centres that use standardised imaging protocols, the protocols should be established
3386   within published DRLs. Until international or national DRLs are established, local DRLs
3387   (LDRLs) should be established as part of the QA programme to inform local policy for
3388   common procedures. LDRLs are established from mean doses delivered to average-sized
3389   patients, with separate LDRLs established for children (IPEM Report 88, 2004). Audits of
3390   standardised protocols should be performed periodically to ensure compliance. Currently,
3391   there is dearth of data on DRLs.

9.5. Audit

3395   (217) Periodic audits of patient imaging studies are recommended to ensure optimal use of
3396   the imaging system. The audit should consider image quality, positioning, FOV, patient
3397   motion, and radiation dose metric. In particular, the audit should evaluate high-dose CBCT
3398   procedures, and repeat CBCT scans. The SEDENTEXCT Consortium report recommends
3399   two audits per year for reject analysis, and a patient dose audit every three years (EC, 2012).

9.6. References

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10. RECOMMENDATIONS

1. Expanded availability and newer applications have put CBCT technology in the hands of medical professionals who traditionally do not use CT. ICRP's radiological protection principles and recommendations as provided in earlier publications, in particular *Publications 87* (Managing patient dose in computed tomography) and *102* (Managing patient dose in multi-detector computed tomography (MDCT)), apply to these newer applications and should be adhered to.

2. Since many applications of CBCT involve patient doses similar to MDCT, the room layout and shielding requirements in such cases need to be similar to adequately protect workers.

3. Medical practitioners bear the responsibility for making sure that each CBCT examination is justified and appropriate.

4. When referring a patient for a diagnostic CBCT examination, the referring practitioner should be aware of the strengths and weaknesses for CBCT *vis-à-vis* MDCT, MRI, and other competing imaging modalities. The decision to utilise CBCT should be made in consultation with imaging professional.

5. Manufacturers are challenged to practice standardised methods for dosimetry and dose display in CBCT in conformance with international recommendations such as ICRU. Unfortunately, at present, there is wide variation in dose quantities being displayed in CBCT machines. Theusers are unable to compare doses among different scanners or protocols.

6. Use of CBCT systems for both fluoroscopy and tomography poses new challenges in quantitating radiation dose. There is a need to develop methods that aggregate exposures to individual patients during the entire procedure that may utilise a combination of fluoroscopy and CBCT during a given examination.

7. Recording, reporting and tracking of radiation dose for a single patient should be made possible.

8. There is a need to provide checks and balances, for example dose check alerts implemented in CT in recent years, to avoid high patient doses as compared to locally defined reference values.

9. Positioning radiosensitive organs such as the thyroid, lens of the eye, breasts and gonads on the detectorside during the partial rotation scan is a useful feature in CBCT that needs to be utilised for radiological protection of these organs.

10. Many machines were initially only capable of fluoroscopy, but can now additionally perform CBCT. Because of the improved clinical information on CBCT, and its ability to remove overlying structures, a user may be tempted to over utilise the CBCT mode. Users must understand that the CBCT function of their system is not a low-dose "fluoroscopy run" and use this mode judiciously.

3489

ANNEX A. ASSESSING PATIENT DOSES IN CBCT

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3491

3492   (A1) This Annex provides a more in-depth description of patient dosimetry methods and
3493 limitations in CBCT. A summarised version is found in Chapter 5. A more extensive
3494 coverage of dosimetry in CBCT is found in ICRU Report 87 (ICRU, 2012).

3495

A.1. Dosimetry in CBCT

3496

3497

3498   (A2) CBCT utilises a wide x-ray beam for 3D imaging of a relatively large volume. Since
3499 the mid-1990s, the trend in MDCT has been towards an ever-increasing number of slices with
3500 a concomitant increase in x-ray beam width; the z-axis coverage of the high-end, wide-area
3501 MDCT scanners available today rivals that of CBCT. These developments have created a
3502 drive to update CT dosimetry methods so that they are more apropos wide area detectors. As
3503 a result, some of the work from MDCT dosimetry, for which established measurement
3504 methods and phantoms already exist, can be translated to CBCT dosimetry. This chapter first
3505 discusses the shortcomings of the standard narrow-beam MDCT formalism when it is directly
3506 applied to CBCT. In order to construct a comprehensive framework for CBCT dosimetry,
3507 methods to overcome these problems are described.

3508   (A3) CT dosimetry has evolved around the concept of the CTDI. From its introduction by
3509 Shope et al. in the 1980's (Shope et al., 1981), CTDI has taken different forms depending on
3510 the adopting organisation: the United States Food and Drug Administration (FDA), the IEC,
3511 and other similar agencies. CTDI has mainly been used to compare dose characteristics of
3512 different CT machines, to test the stability of equipment performance (quality control), and,
3513 in some instances, to estimate patient dose even though CTDI does not directly provide an
3514 assessment of patient dose. An extensive description of the CTDI concept is found in ICRU
3515 Report 87 (ICRU, 2012).

3516   (A4) Increasingly, wide beams in modern CT and CBCT scanners complicate CTDI
3517 measurements. Even for a nominal beam width of 20 mm, it is evident that the 100-mm
3518 typical chamber cannot collect the tails of the dose profile in a poly(methyl methacrylate)
3519 (PMMA) phantom. The ratio of $CTDI_{100}/CTDI_{\infty}$ is called CTDI measurement efficiency.
3520 Kyriakou et al. (2008) have shown that for a 200-mm collimation, an integration length of
3521 >600 mm would be required to approximate $CTDI_{\infty}$ within 1%.

3522   (A5) This definition of efficiency has been the basis of the new approach of wide-beam CT
3523 dosimetry. The IAEA (2011) adopted a two-step approach proposed by the IEC (2010). More
3524 details regarding this modified approach are found in ICRU Report 87 (ICRU, 2012).

3525   (A6) It would be useful to mention that CTDI alone is not a useful indicator of patient dose.
3526 In order to connect the CTDI-like measurements with dose, $CTDI_{vol}$ and DLP have been
3527 extensively used in clinical practice as relative patient dose indicators. $CTDI_{vol}$ and DLP are
3528 connected by the equation:

$$DLP = L \cdot CTDI_{vol}$$

3529

3530 where L is the length of the scan. As discussed later on in this annex, the $CTDI_{vol}$ paradigm is
3531 problematic in cases where there is no helical scan or patient motion (as is the case with many
3532 CBCT scanners). In such cases, reported $CTDI_{vol}$ values will significantly overestimate the
3533 dose (Dixon and Boone, 2010a).

3534

A.2. Point of care scanning and physicians clinic based CBCT systems

3535

3536

3537   (A7) Clinic-based systems include head and neck CBCT, bCT and dental CBCT. One of
3538   the main differences between dental and other clinic-based scanners (i.e. head and neck
3539   scanners) is the FOV, as head and neck scanners are capable of imaging larger volumes.

3540   (A8) For dental systems, the SEDENTEXCT Consortium report (EC, 2012) discussed the
3541   use of KAP as well as CTDI-like measurements. It was proposed that CTDI measurements
3542   should be carried out during commissioning in cases when the machine comes with data on
3543   such measurements from the manufacturer. On the grounds that the conventional CTDI has
3544   drawbacks for dental CBCT use (due to wider beams and greater asymmetry of dose
3545   distribution in CBCT compared to MDCT), the consortium tried to define a single CBCT DI
3546   (Pauwels et al., 2012). During this effort, a customised phantom (SEDENTEXCT DI) was
3547   developed in collaboration with Leeds Test Objects Ltd (Boroughbridge UK) which is shown
3548   in Fig. A.1. It features suitable insets for the placement of measuring equipment. The
3549   phantom consists of four ionisation chamber plates (2 x 22 mm and 2 x 44 mm), one TLD
3550   plate (22 mm thick), and one film plate (22 mm thick). Three adapters with widths of 22, 44
3551   and 66 mm are provided that can reduce the chamber diameter from 26 to 13 mm. Two
3552   different measurement setups (Index 1 and Index 2) are depicted in Fig. A.1. Index 1 is
3553   suitable for assessment of dose distribution for on-axis and off-axis exposures by rotating the
3554   phantom so that the beam isocentre lies on the diameter of the phantom. Index 2 is suitable
3555   for measuring symmetric dose distributions. Measurements are taken on the central axis of
3556   the phantom and at peripheral positions near the surface of the phantom. Pauwels et al. (2012)
3557   concluded that there is no optimal dose index for dental CBCT mostly due to the complicated
3558   geometry and practical aspects of the quality control measurements. Further validation of
3559   possible indices is required together with a way to translate dose index' readings into patient
3560   doses. Araki et al. (2013) concluded that CBCT DI and KAP proposed by SEDENTEXCT
3561   could be used to establish DRLs for dental CBCT. The same authors note that the relationship
3562   of these indices to effective dose remains to be determined.

3563
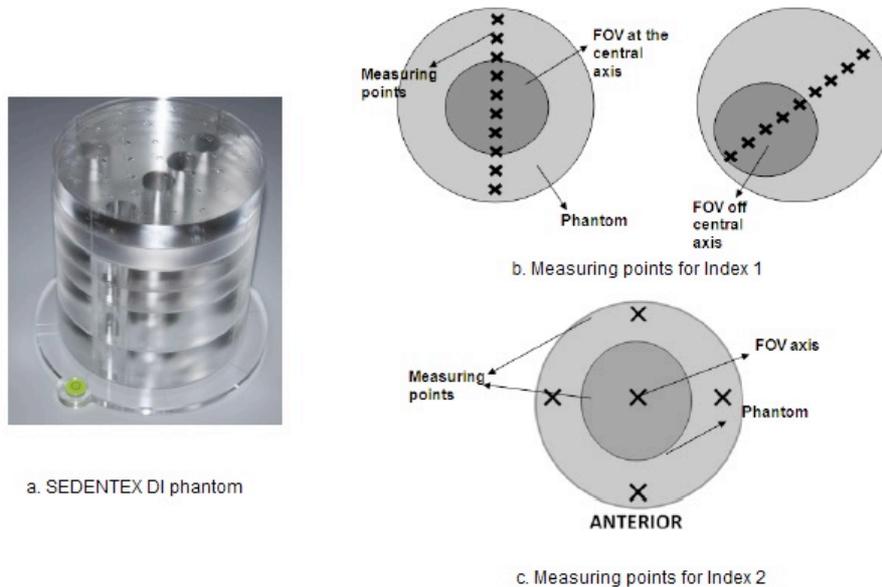
3564
3565   Fig. A.1. (a) The SEDENTEX DI phantom for radiation dose measurements in dental CBCT
3566   systems. (b) and (c) Measuring points for the estimation of index 1 and 2. The DI phantom
3567   allows for seven measurements for index 1. (permissions required)

3568

3569   (A9) It has been suggested that if the manufacturer has provided a CTDI dose figure, then
3570   this quantity should be measured during commissioning. However, not all machines come

3571   with such initial measurements. Another dose index used for CBCT dosimetric evaluations is
3572   the KAP which is often used in panoramic and cephalometric radiography and, of course, is
3573   widely used in radiography and fluoroscopy. Some machines display a KAP value on screen
3574   after the exposure. The accuracy of such measurements should be verified by medical
3575   physicists. The use of KAP has been proposed by the UK HPA (2010a) currently named
3576   Public Health England. The main advantage of KAP is that it is easy to calculate by
3577   measuring dose and beam cross-section at a specific point. It is considered suitable for
3578   auditing CBCT dose in dental practices (HPA, 2010b). The SEDENTEXCT Consortium
3579   proposes that if such measurements are not provided, the medical physicist should create a
3580   log of such readings in all clinically used settings so that the dentist may compare with
3581   national and international audit levels (EC, 2012).

3582   (A10) Technically the methods described above could also be applied to other clinic-based
3583   systems including, for example, systems for head and neck imaging and possibly bCT.
3584   However, there is currently no standardisation in the measurements for such units. This
3585   highlights more vividly that the issue of standardisation in CBCT dosimetry remains largely
3586   unresolved.

A.3. C-arm CBCT systems

3590   (A11) C-arm CBCT systems are incapable of performing a full rotation around the patient
3591   couch. Some systems can only rotate 180° plus the beam angle (Fahrig et al., 2006), which
3592   results in a non-uniform axial dose deposition to the patient/phantom. In a phantom, the
3593   maximum dose occurs at the central plane intersecting the z-axis at z = 0, on the side of the
3594   phantom closest to the x-ray tube. In the ideal case in which the heel effect is absent, the
3595   maximum dose would occur on the bisector of the rotation angle. When the heel effect is
3596   present, the maximum dose occurs near the bisector.

3597   (A12) For C-arm CBCT systems, Fahrig et al. (2006) proposed a metric representing the
3598   average dose to the phantom central plane (z = 0)

$$\bar{D}(0) = \frac{1}{3}D_0 + \frac{2}{3}\bar{D}_p$$

3600   where D_0 is the dose to the central point of the central plane (on the z-axis) and D_p is the
3601   average peripheral dose. This equation follows a similar averaging to that used in the
3602   calculation of the $CTDI_w$, the metric that is used for dosimetry on any conventional CT
3603   scanner performing a rotation smaller than 360°. Fahrig et al. (2006) performed the
3604   calculation using a Farmer ionisation chamber and measured doses at the centre and at eight
3605   peripheral positions at 1 cm depth from the head phantom's surface. Podnieks and Negus
3606   (2012) showed that effective dose can be estimated from the $CTDI_w$ and the irradiated length
3607   to an acceptable accuracy if the ionisation chamber positions are considered carefully.

A.4. A unified approach to CT dosimetry

3611   (A13) The ICRU (2012) in its Report 87 has reviewed a considerable body of work in order
3612   to propose a method for CT dosimetry that compensates for the shortcomings of current
3613   CTDI-based CT dosimetry methods. In addition, earlier work by Dixon and Boone (2010b)
3614   provided a unified formalism for dose measurements on machines capable of helical scanning
3615   (e.g. MDCTs) as well as on those that only acquire axial images (which is the case with most
3616   CBCTs). A set of metrics and the use of a new polyethylene 600-mm long phantom are
3617   proposed. This method has previously been described in AAPM Report 111 (AAPM, 2010),
3618   but in this publication, the notation as presented in ICRU Report 87 was used. The

3619   mathematical foundation for the method is beyond the scope of this publication; however, the
3620   method is briefly discussed below.

3621   (A14) A dosimetry quantity $CTDI_L$ is proposed, the physical meaning of which is the dose
3622   at the centre (z = 0) of the scanned length for a scan from a z = −L/2 to z = L/2. This
3623   formalism provides a means to estimate the dose deposited at the central plane of the phantom,
3624   at z = 0. In the case of axial scans, such as those performed with most CBCT machines, the
3625   quantity that intuitively corresponds to the CTDI is the dose at the central point of the beam
3626   on the z-axis. If f(z) is the dose profile function, then this dose is in fact f(0). For a number of
3627   N identical axial scans centred at z = 0, the dose of interest will be equal to Nf(0).

3628
3629   **A.4.1. Formalism**
3630

3631   (A15) For a helical CT scan, the accumulated absorbed dose distribution at the centre of the
3632   scan length (from –L/2 to +L/2) is represented by a convolution of the axial dose profile with
3633   a rectangular function, $\Pi(z/L)$ of scan length L. This representation is only valid when x-ray
3634   tube current modulation is not used. Fig. A.2. shows normalised cumulative absorbed dose
3635   distributions for a series of helical CT scans of differing scan lengths, produced by Monte
3636   Carlo simulation (Boone, 2009).

3637
3638
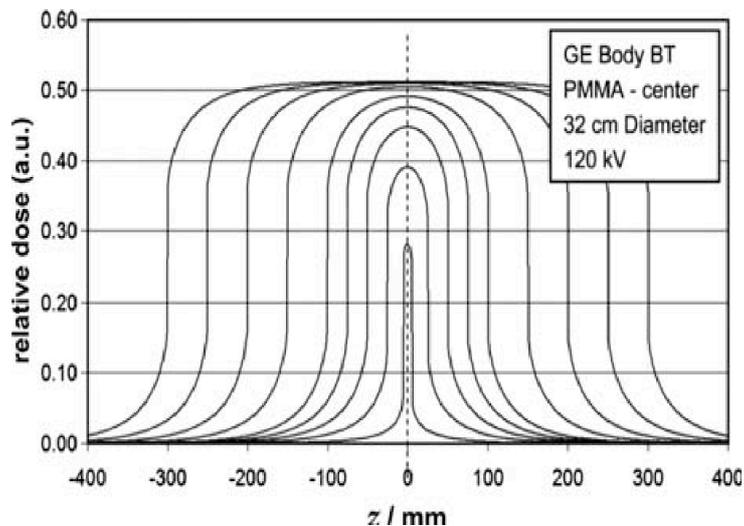
3639
3640   Fig. A.2. Normalised absorbed dose as a function of z-position for a number of different scan
3641   lengths: 10 mm, 50 mm, 100 mm, 150 mm, 200 mm, 300 mm, 400 mm, 500 mm, and 600
3642   mm (from centre to edge on the graph). These data were derived by convolving the dose
3643   spread function (DSF) computed from the Monte Carlo simulation with rectangular functions
3644   characterising the length of the scan, for a 320-mm diameter PMMA phantom at 120 kV,
3645   using a GE Lightspeed 16 body bowtie filter. (Source: ICRU, 2012). (permissions required)

3646
3647   (A16) The dose $D_L(0)$ at the central part of the beam (z = 0) for a beam width L, increases
3648   as the width of the beam increases. This can be seen in Fig. A.2. $D_L(0)$ approaches
3649   asymptotically a maximum value when the beam width increases. This value is called the
3650   equilibrium dose (D_{eq}). This value could be understood as the $CTDI_\infty$, i.e. when the entire
3651   dose profile has been collected.

3652
3653   **A.4.2. Cumulative absorbed dose distribution from a helical scan of length L**
3654

3655   (A17) The cumulative absorbed dose distribution $D_L(z)$ for helical scans in which the table
3656   moves by a distance b per gantry rotation, can be calculated by using the following equation
3657   which is only applicable when tube current modulation is not used

$$D_L(z) = \frac{1}{b} \int_{-L/2}^{+L/2} f(z - z') \, dz'$$

3658
3659   (A18) At z = 0 and taking into account that pitch (p) is defined as p = b/nT, the above
3660   equation becomes

$$D_L(0) = \frac{1}{b} \int_{-L/2}^{+L/2} f(z')$$

3661
3662

$$dz' = p \cdot CTDI_L$$

3663
3664   (A19) Note that for p = 1, $D_L(0)$ = $CTDI_L$. Conceptually, $D_L(0)$ as a function of L uses the
3665   data points along a vertical line perpendicular to z = 0 in Fig. A.2.
3666   (A20) $D_L(0)$ depends on L, until the asymptote D_{eq} is reached at very long scan lengths. A
3667   new function capable of representing this dependence needs to be introduced. The
3668   mathematical synonym function $h(L) = D_L(0)$ is thus the following:

$$h(L) = \frac{1}{b} \int_{-L/2}^{+L/2} f(z') \, dz'$$

3669
3670   (A21) Conceptually, h(L) is the integral of the intercepted dose profile on the z axis for a
3671   scan of length L by keeping the detector at the centre of the phantom.
3672   (A22) If the cumulative absorbed dose at z = 0 is normalised to D_{eq}, the above equation
3673   becomes
3674

$$H(L) = \frac{h(L)}{D_{eq}} = \frac{D_L(0)}{D_{eq}}$$

3675
3676   (A23) Fig. A.3. shows H(L) curves measured by Mori et al. (2005). The maximum H(L)
3677   value as a function of scan length L asymptotically approaches unity for large scan lengths.
3678   This has been referred to as the rise to dose equilibrium curve. Because H(L) is normalised to
3679   unity at L→∞, this function does not contain the tube output information that h(L) does.
3680
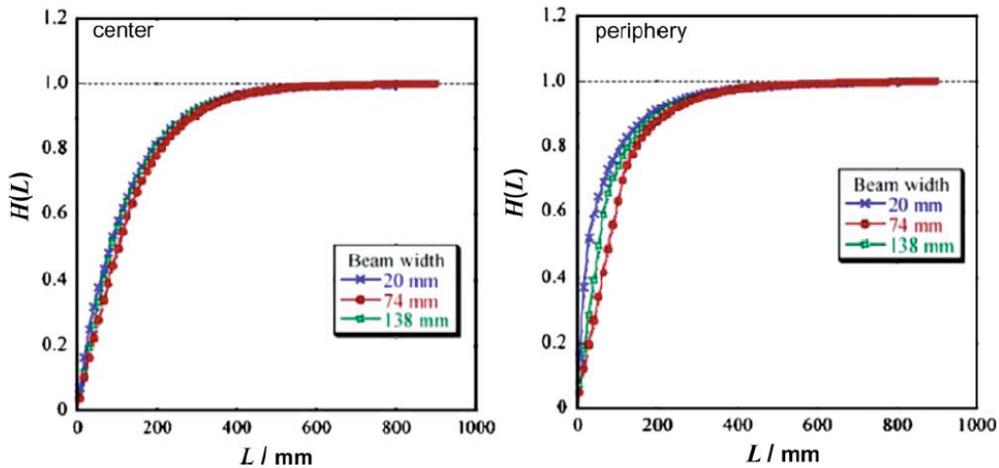
3681

3682 Fig. A.3. Graphs showing measured H(L) curves. These data were measured in a 900-mm
3683 long, 320-mm-diameter PMMA phantom, scanned at 120 kV. Three different beam
3684 collimation widths are shown in each plot, for the (a) centre and (b) periphery positions.
3685 (Source: Mori et al., 2005). (permissions required)
3686
3687   (A24) The physical interpretation of the rise to equilibrium curve is that the scan and the
3688 phantom need to be long enough so that the asymptote tails of the profiles are reached. The
3689 longer the scan, the more H(L) approaches unity. This representation is therefore good in
3690 showing the relatively low efficiency of short scans for collecting the actual dose, and this
3691 efficiency increases with longer scans.
3692
3693 **A.4.3. Phantoms**
3694
3695   (A25) It has been shown that a phantom with a 300-mm diameter would need to be at least
3696 400 mm in length to capture ~98% of D_{eq} (this is equivalent to saying that the scan profile
3697 interception would be 98% efficient). For a phantom with the standard 320 mm diameter, a
3698 length of 425 mm would be required for the same measurement efficiency. To tackle this
3699 problem, the committee responsible for ICRU Report 87 collaborated with the AAPM task
3700 group responsible for the upcoming Report 200. As a result of this collaboration, the phantom,
3701 ICRU/AAPM TG 200, shown in Fig. A.4 was developed.
3702
3703
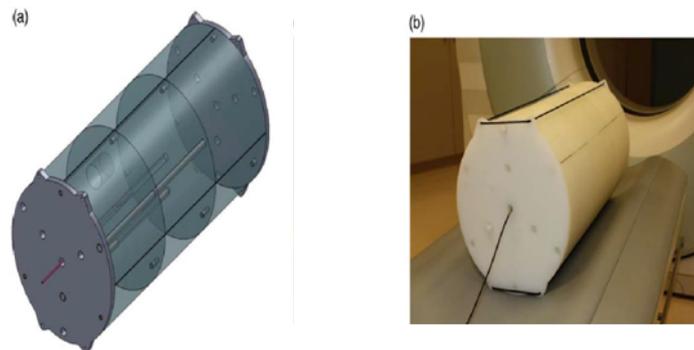
3704
3705 Fig. A.4. The ICRU/AAPM TG 200 phantom. The phantom is made of high density
3706 polyethylene (0.97 g/cm^3). With a diameter of 300 mm and a length of 600 mm, which are
3707 sufficient for measuring functions, h(L) or H(L). Panel (a) illustrates the design of this
3708 phantom, and panel (b) shows a photograph of the phantom. The phantom is large and weighs
3709 about 41 kg. Therefore, it was designed to be modular, with three different sections. (Source:
3710 ICRU, 2013). (permissions required)
3711
3712 **A.4.4. Practical measurement of rise-to-equilibrium dose curves**
3713
3714   (A26) Methods for measuring the H(L) or h(L) curves have been described in adequate
3715 extent in AAPM Report 111 (2010) and ICRU Report 87 (2012). Here, a short and intuitive
3716 description of the measurement methods is given.
3717   (A27) A long phantom and an integrating thimble ionisation chamber are needed. A series
3718 of helical scans of different lengths is performed, and the air kerma integrated by the thimble
3719 chamber is recorded. The scans are centred on the position of the chamber. The air kerma
3720 readings as measured by the chamber are plotted as a function of length of the helical scan.

(A28) If a real-time radiation dosimeter is available, the rise-to-equilibrium curve may be plotted using data obtained during a single long scan. In this case, the dosimeter can create a full dose profile along the whole length of the phantom. Different points on the curve may then be calculated by integrating the dose profile curve using appropriate integration limits (– L/2 to L/2), where L is the total integration length centred on the real time radiation meter at the centre of the phantom.

A.4.5. Measurements on machines only capable of axial acquisition

(A29) The methods described above are useful for measurements in MDCT machines that provide the option to perform helical scans. However, some CBCT machines may not perform helical scans. When table translation during a scan is not available, it is necessary to modify the method, based on the notion that it is necessary to measure a quantity that corresponds to the CTDI of helical scans. As mentioned previously, this quantity is f(0) (Dixon and Boone, 2010b). Practically speaking, f(0) is measured by placing the ionisation chamber at the centre of the phantom and the beam and varying the beam width starting from the thinnest possible collimation to the widest available. The measurement values can then be plotted against the beam width α. The values may be normalised to A_{eq} which is the equilibrium value that would be reached for f(0) if the beam width was $\geq$470 mm. Such beam widths are, of course, not found in clinical practice. Thus, the normalised approach-to-equilibrium-curve for the axial scan is only partial, and does not asymptotically reach the value of 1. For axial CT scans with a cone beam width α, dose $f(0)_\alpha = H(\alpha)A_{eq}$, the conventional CT dose $D_L(0)$ can be described as a function of scan length L, including a common equilibrium dose constant A_{eq}, a common scatter equilibrium length $\alpha_{eq} = L_{eq}$, and a common function $H(\lambda)$ which describes the relative approach to dose equilibrium for both modalities, where $\lambda = \alpha$, or $\lambda = L$, such that $f(0)_\alpha = H(\alpha)A_{eq}$ and $D_L(0) = H(L)D_{eq} = H(L)(b/\alpha)A_{eq}$. Axial scanners that do not have the facility to collimate the beam, may be equipped with a collimation gauge that could be inserted before the x-ray tube for dose measurement purposes.

(A30) It is important to note here that the integration which needs to be performed in order to measure CTDI is a result of the existence of table movement. The definition of CTDI implies that dose to the central area of a phantom is affected by scatter from adjacent areas. This phenomenon is completely absent in axial scans, and therefore, CTDI consistently overestimates the dose around the central area of the phantom.

A.4.6. ICRU Report 87 recommendations

CTDI$_{vol}$ and CTDI$_{air}$ measurements

(A31) CTDI$_{vol}$ has been traditionally related to measurements of CT dose. The IEC has also recommended that CTDI$_{vol}$ be displayed on the control screen of CT scanners. Due to its widespread use and in order to keep continuity with older measurements on CT scanners, the ICRU recommends that CTDI$_{vol}$ as well as CTDI$_{vol}$ free-in-air be measured at acceptance testing using both 160-mm and 320-mm diameter PMMA phantoms, at clinically relevant mAs settings across the range of clinically used tube potentials. Furthermore, CTDI$_{vol}$ is used to scale size-specific dose estimates (SSDE) as well as for normalisation of rise-to-equilibrium curves. The x-ray output of the CT scanner, which is also characterised by CTDI$_{air}$, is a fundamental measurement that should be performed during acceptance testing and after changing major components of the scanner related to dose.

Dosimetry in phantoms

3771 (A32) If medical physicists follow the recommendations and measure $CTDI_{vol}$ and $CTDI_{air}$ at acceptance testing, measurements of $CTDI_{vol}$ in phantoms should not be needed on a routine basis if periodic $CTDI_{air}$ measurements are stable.

(A33) Manufacturers should measure and provide users with a comprehensive set of data for a reasonably wide range of beam settings used in clinical practice regarding the rise-to-equilibrium curves of the scanner and related metrics such as H(L) and h(L). G(L) which is the H(L) curve normalised by $CTDI_{vol}$ and thus related to patient dose should also be provided.

(A34) A subset of CTDI measurements performed by only using the central 200-mm section of the phantom should also be provided by manufacturers so that G(L) measured for the full 600-mm phantom can be associated to the partial G(L) measurement acquired with the 200-mm phantom section.

Patient dose estimations

(A35) Patient dose can be estimated by using SSDE coupled with the $CTDI_{vol}$. The method has been described in ICRU Report 87 (2012) and AAPM Report 204 (2011). It must be considered, however, that $CTDI_{vol}$ calculation can be different for partial rotation axial CT scans, such as in the case of C-arm CBCT scan. Even for full axial scans in which there is no patient translation, the $CTDI_{vol}$ will overestimate patient dose (Dixon and Boone, 2010b). This fact underlines the need for new coefficients for patient dose estimation from f(0) measurements.

A.5. Tracking and reporting of radiation dose

(A36) New challenges emerge with systems being used for both fluoroscopy and tomography (CBCT). While fluoroscopy radiation dose figures are normally available as KAP from the machines, CBCT doses are currently provided by different manufacturers in different units. Currently, there is no way to assess the aggregate radiation dose to a patient during a single procedure. Further, there is a need to facilitate comparison of radiation doses to patients between a single run of CT to one or several DSA series. This situation needs to be addressed, and a system should provide a means of not only comparing but also consolidating doses from both fluoroscopy and CT. Furthermore, tracking and reporting of radiation dose for a single patient should be made possible, as it is becoming increasingly important to do this for strengthening the processes involved in the justification and optimisation principles of ICRP (Rehani and Frush, 2011; Seuri et al., 2013).

A.6. Epilogue

(A37) Different methods for CBCT dosimetry have been presented. However, in order to be able to evaluate CBCT usefulness in regard to its alleged dose reduction in comparison to CT, a metric which could be used for direct comparisons is needed. The unified CT dosimetry method proposed by ICRU (2012) has the potential to standardise CBCT dosimetry. This method can be implemented without updating the equipment already in use in the clinical CT arena. Furthermore, the methods discussed could be used to measure dose for many types of different CBCT systems, including radiotherapy CBCT, clinic-based systems, dedicated breast systems, and C-arm systems. The value of CTDI-based measurements also presented in this chapter should not be underestimated. Although CTDI has limitations, it has been evaluated on many systems over the years and provides important comparisons in output for

CT scanners from different manufacturers and ages. Also the coefficients for patient dose estimations that are available today are based on the $CTDI_{vol}$.

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