

## What is Cone-Beam CT and How Does it Work?

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Imaging is an important diagnostic adjunct to the clinical assessment of the dental patient. The introduction of panoramic radiography in the 1960s and its widespread adoption throughout the 1970s and 1980s heralded major progress in dental radiology, providing clinicians with a single comprehensive image of jaws and maxillofacial structures. However, intraoral and extraoral procedures, used individually or in combination, suffer from the same inherent limitations of all planar two-dimensional (2D) projections: magnification, distortion, superimposition, and misrepresentation of structures. Numerous efforts have been made toward three-dimensional (3D) radiographic imaging (eg, stereoscopy, tuned aperture CT) and although CT has been available, its application in dentistry has been limited because of cost, access, and dose considerations. The introduction of cone-beam computed tomography (CBCT) specifically dedicated to imaging the maxillofacial region heralds a true paradigm shift from a 2D to a 3D approach to data acquisition and image reconstruction. Interest in CBCT from all fields of dentistry is unprecedented because it has created a revolution in maxillofacial imaging, facilitating the transition of dental diagnosis from 2D to 3D images and expanding the role of imaging from diagnosis to image guidance of operative and surgical procedures by way of third-party applications software.

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The purpose of this article is to provide an overview of this CBCT technology and an understanding of the influence of technical parameters on image quality and resultant patient radiation exposure.

## Background

CBCT is a recent technology. Imaging is accomplished by using a rotating gantry to which an x-ray source and detector are fixed. A divergent pyramidal- or cone-shaped source of ionizing radiation is directed through the middle of the area of interest onto an area x-ray detector on the opposite side. The x-ray source and detector rotate around a rotation fulcrum fixed within the center of the region of interest. During the rotation, multiple (from 150 to more than 600) sequential planar projection images of the field of view (FOV) are acquired in a complete, or sometimes partial, arc. This procedure varies from a traditional medical CT, which uses a fan-shaped x-ray beam in a helical progression to acquire individual image slices of the FOV and then stacks the slices to obtain a 3D representation. Each slice requires a separate scan and separate 2D reconstruction. Because CBCT exposure incorporates the entire FOV, only one rotational sequence of the gantry is necessary to acquire enough data for image reconstruction (Fig. 1).

CBCT was initially developed for angiography [1], but more recent medical applications have included radiotherapy guidance [2] and mammography [3]. The cone-beam geometry was developed as an alternative to conventional CT using either fan-beam or spiral-scan geometries, to provide more rapid acquisition of a data set of the entire FOV and it uses a comparatively less expensive radiation detector. Obvious advantages of such a system, which provides a shorter examination time, include the reduction of image unsharpness caused by the translation of the patient, reduced image distortion due to internal patient movements, and increased x-ray tube efficiency. However, its main disadvantage, especially with larger FOVs, is a limitation in image quality related to noise and contrast resolution because of the detection of large amounts of scattered radiation.

It has only been since the late 1990s that computers capable of computational complexity and x-ray tubes capable of continuous exposure have enabled clinical systems to be manufactured that are inexpensive and small enough to be used in the dental office. Two additional factors have converged to make CBCT possible.

### *Development of compact high-quality two-dimensional detector arrays*

The demands on any x-ray detector in clinical CBCT are hard to fulfill. The detector must be able to record x-ray photons, read off and send the signal to the computer, and be ready for the next acquisition many hundreds of times within a single rotation. Rotation is usually performed within times equivalent to, or less than, panoramic radiography (10–30 seconds), which

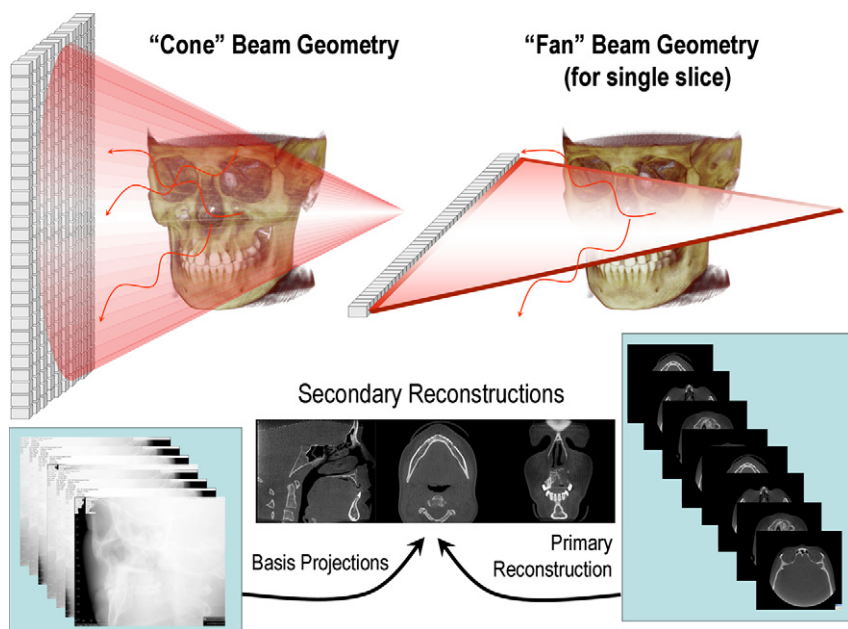


Fig. 1. X-ray beam projection scheme comparing acquisition geometry of conventional or “fan” beam (*right*) and “cone” beam (*left*) imaging geometry and resultant image production. In cone-beam geometry (*left*), multiple basis projections form the projection data from which orthogonal planar images are secondarily reconstructed. In fan beam geometry, primary reconstruction of data produces axial slices from which secondary reconstruction generates orthogonal images. The amount of scatter generated (sinusoidal lines) and recorded by cone-beam image acquisition is substantially higher, reducing image contrast and increasing image noise.

necessitates frame rate image acquisition times of milliseconds. Detectors were initially produced using a configuration of scintillation screens, image intensifiers, and charge-coupled device (CCD) detectors. However, image intensifier systems are large and bulky and FOVs may suffer from peripheral truncation effects (volumetric “cone cuts”), having circular entrance areas rather than more appropriate rectangular ones. Furthermore, rotation of the source-to-detector arrangement may influence sensitivity because of the interference between the magnetic field of the earth and those in the image intensifiers. More recently, high-resolution, inexpensive flat-panel detectors have become available. Such flat detectors are composed of a large-area pixel array of hydrogenated amorphous silicon thin-film transistors. X rays are detected indirectly by means of a scintillator, such as terbium-activated gadolinium oxysulphide or thallium-doped cesium iodide, which converts X rays into visible light that is subsequently registered in the photo diode array. The configuration of such detectors is less complicated and offers greater dynamic range and reduced peripheral distortion; however, these detectors require a slightly greater radiation exposure.

### *Refinement of approximate cone-beam algorithms*

Reconstructing 3D objects from cone-beam projections is a fairly recent accomplishment. In conventional fan-beam CT, individual axial slices of the object are sequentially reconstructed using a well-known mathematic technique (filtered back projection) and subsequently assembled to construct the volume. However, with 2D x-ray area detectors and cone-beam geometry, a 3D volume must be reconstructed from 2D projection data, which is referred to as “cone-beam reconstruction.” The first and most popular approximate reconstruction scheme for cone-beam projections acquired along a circular trajectory is the algorithm according to Feldkamp and colleagues [4], referred to as the Feldkamp, Davis, and Kress (FDK) method. This algorithm, used by most research groups and commercial vendors for CBCT with 2D detectors, uses a convolution-back projection method. Although it can be implemented easily with currently available hardware and is a good reconstruction for images at the center or “midplane” of the cone beam, it provides an approximation that causes some unavoidable distortion in the noncentral transverse planes, and resolution degradation in the longitudinal direction. To address this deficiency, several other approaches have been proposed using different algorithms [5] and cone-beam geometries (eg, dual orthogonal circles, helical orbit, orthogonal circle-and-line), and these will no doubt be incorporated into future CBCT designs.

### **Cone-beam CT image production**

Current cone-beam machines scan patients in three possible positions: (1) sitting, (2) standing, and (3) supine. Equipment that requires the patient to lie supine physically occupies a larger surface area or physical footprint and may not be accessible for patients with physical disabilities. Standing units may not be able to be adjusted to a height to accommodate wheelchair-bound patients. Seated units are the most comfortable; however, fixed seats may not allow scanning of physically disabled or wheelchair-bound patients. Because scan times are often greater than those required for panoramic imaging, perhaps more important than patient orientation is the head restraint mechanism used. Despite patient orientation within the equipment, the principles of image production remain the same.

The four components of CBCT image production are (1) acquisition configuration, (2) image detection, (3) image reconstruction, and (4) image display. The image generation and detection specifications of currently available systems (Table 1) reflect proprietary variations in these parameters.

### *Acquisition configuration*

The geometric configuration and acquisition mechanics for the cone-beam technique are theoretically simple. A single partial or full rotational

Table 1  
Selected CBCT imaging systems.

Unit	Model(s)	Manufacture/distributor
Accuitomo	3D Accuitomo - XYZ Slice View Tomograph/ Veraviewpacs 3D	J. Morita Mfg. Corp., Kyoto, Japan
Galileos	—	Sirona Dental Systems, Charlotte, North Carolina
Hitachi	CB MercuRay/CB Throne	Hitachi Medical Systems, Tokyo, Japan
i-CAT	Classic/Next Generation	Imaging Sciences International, Hatfield, Pennsylvania
ILUMA	Ultra Cone Beam CT Scanner	IMTEC Imaging, Ardmore, Oklahoma; distributed by KODAK Dental Systems, Carestream Health, Rochester, New York
KaVo	3D eXam	KaVo Dental Corp., Biberach, Germany
KODAK	9000 3D	KODAK Dental Systems, Carestream Health, Rochester, New York
NewTom	3G/NewTom VG	QR, Inc., Verona, Italy/Dent-X Visionary Imaging, Elmsford, New York
Picasso Series	Trio/Pro/Master	E-Woo Technology Co., Ltd./Vatech, Giheung-gu, Korea
PreXion 3D	—	TeraRecon Inc., San Mateo, California
Promax	3D	Planmeca OY, Helsinki, Finland
Scanora	3D Dental conebeam	SOREDEX, Helsinki, Finland
SkyView	3D Panoramic imager	My-Ray Dental Imaging, Imola, Italy

scan from an x-ray source takes place while a reciprocating area detector moves synchronously with the scan around a fixed fulcrum within the patient's head.

### *X ray generation*

During the scan rotation, each projection image is made by sequential, single-image capture of attenuated x-ray beams by the detector. Technically, the easiest method of exposing the patient is to use a constant beam of radiation during the rotation and allow the x-ray detector to sample the attenuated beam in its trajectory. However, continuous radiation emission does not contribute to the formation of the image and results in greater radiation exposure to the patient. Alternately, the x-ray beam may be pulsed to coincide with the detector sampling, which means that actual exposure time is markedly less than scanning time. This technique reduces patient radiation dose considerably. Currently, four units (Accuitomo, CB MercuRay, Iluma Ultra Cone, and PreXion 3D) provide continuous radiation exposure. Pulsed x-ray beam exposure is a major reason for considerable variation in reported cone-beam unit dosimetry.

### *Field of view*

The dimensions of the FOV or scan volume able to be covered depend primarily on the detector size and shape, the beam projection geometry,

and the ability to collimate the beam. The shape of the scan volume can be either cylindric or spherical (eg, NewTom 3G). Collimation of the primary x-ray beam limits x-radiation exposure to the region of interest. Field size limitation therefore ensures that an optimal FOV can be selected for each patient, based on disease presentation and the region designated to be imaged. CBCT systems can be categorized according to the available FOV or selected scan volume height as follows:

Localized region: approximately 5 cm or less (eg, dentoalveolar, temporomandibular joint)

Single arch: 5 cm to 7 cm (eg, maxilla or mandible)

Interarch: 7 cm to 10 cm (eg, mandible and superiorly to include the inferior concha)

Maxillofacial: 10 cm to 15 cm (eg, mandible and extending to Nasion)

Craniofacial: greater than 15 cm (eg, from the lower border of the mandible to the vertex of the head)

Extended FOV scanning incorporating the craniofacial region is difficult to incorporate into cone-beam design because of the high cost of large-area detectors. The expansion of scan volume height has been accomplished by one unit (iCAT Extended Field of View model) by the software addition of two rotational scans to produce a single volume with a 22-cm height. Another novel method for increasing the width of the FOV while using a smaller area detector, thereby reducing manufacturing costs, is to offset the position of the detector, collimate the beam asymmetrically, and scan only half the patient (eg, Scanora 3D, SOREDEX, Helsinki, Finland) (Fig. 2).

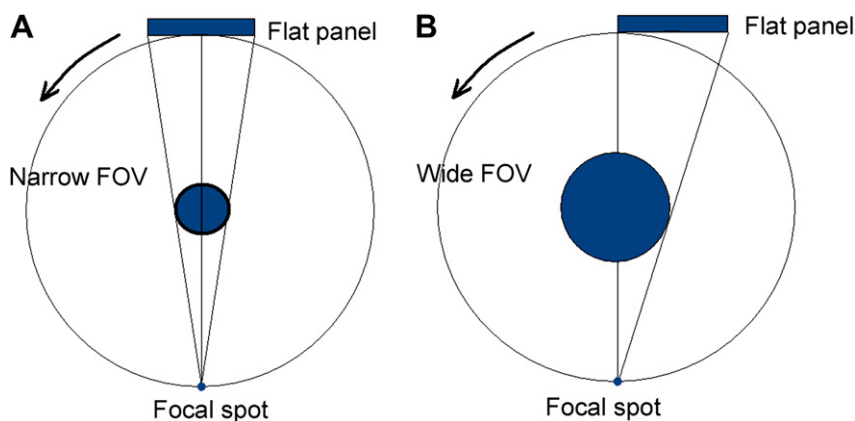


Fig. 2. Novel method of acquiring an extended FOV using a flat panel detector. (A) Conventional geometric arrangement whereby the central ray of the x-ray beam from the focal source is directed through the middle of the object to the center of the flat panel detector. (B) Alternate method of shifting the location of the flat panel imager and collimating the x-ray beam laterally to extend the FOV object. (Courtesy of SOREDEX, Helsinki, Finland; with permission.)

### *Scan factors*

During the scan, single exposures are made at certain degree intervals, providing individual 2D projection images, known as “basis,” “frame,” or “raw” images. These images are similar to lateral and posterior-anterior “cephalometric” radiographic images, each slightly offset from one another. The complete series of images is referred to as the “projection data.” The number of images comprising the projection data throughout the scan is determined by the frame rate (number of images acquired per second), the completeness of the trajectory arc, and the speed of the rotation. The number of projection scans comprising a single scan may be fixed (eg, NewTom 3G, Iluma, Galileos, or Promax 3D) or variable (eg, i-CAT, PreXion 3D). More projection data provide more information to reconstruct the image; allow for greater spatial and contrast resolution; increase the signal-to-noise ratio, producing “smoother” images; and reduce metallic artifacts. However, more projection data usually necessitate a longer scan time, a higher patient dose, and longer primary reconstruction time. In accordance with the “as low as reasonably achievable” (ALARA) principle, the number of basis images should be minimized to produce an image of diagnostic quality.

*Frame rate and speed of rotation.* Higher frame rates provide images with fewer artifacts and better image quality [6]. However, the greater number of projections proportionately increases the amount of radiation a patient receives. Detector pixels must be sensitive enough to capture radiation adequate to register a high signal-to-noise output and to transmit the voltage to the analog and the digital converter, all within a short arc of exposure. Within the limitations of solid-state detector readout speed and the need of short scanning time in a clinical setting, the total number of available view angles is normally limited to several hundred.

*Completeness of the trajectory arc.* Most CBCT imaging systems use a complete circular trajectory or a scan arc of 360° to acquire projection data. This physical requirement is usually necessary to produce projection data adequate for 3D reconstruction using the FDK algorithm (see section on reconstruction). However, it is theoretically possible to reduce the completeness of the scanning trajectory and still reconstruct a volumetric data set. This approach potentially reduces the scan time and is mechanically easier to perform. However, images produced by this method may have greater noise and suffer from reconstruction interpolation artifacts. Currently, this technique is used by at least two units (Galileos and Promax 3D).

### **Image detection**

Current CBCT units can be divided into two groups, based on detector type: an image intensifier tube/charge-coupled device (IIT/CCD) combination or a flat-panel imager.

The IIT/CCD configuration comprises an x-ray IIT coupled to a CCD by way of a fiber optic coupling. Flat-panel imaging consists of detection of X rays using an “indirect” detector based on a large-area solid-state sensor panel coupled to an x-ray scintillator layer. Flat-panel detector arrays provide a greater dynamic range and greater performance than the II/CCD technology. Image intensifiers may create geometric distortions that must be addressed in the data processing software, whereas flat-panel detectors do not suffer from this problem. This disadvantage could potentially reduce the measurement accuracy of CBCT units using this configuration. II/CCD systems also introduce additional artifacts [7].

CBCT systems that use flat-panel detectors also have limitations in their performance that are related to linearity of response to the radiation spectrum, uniformity of response throughout the area of the detector, and bad pixels. The effects of these limitations on image quality are most noticeable at lower and higher exposures. To overcome this problem, detectors are linearized piecewise and exposures that cause nonuniformity are identified and calibrated. In addition, pixel-by-pixel standard deviation assessment is used in correcting nonuniformity. Bad pixels are also examined and most often replaced by the average of the neighboring pixels.

A reduction in image matrix size is desirable to increase spatial resolution and therefore provide greater image detail. However, detector panels comprise an array of individual pixels with two components, photodiodes that actually record the image and thin-film transistors that act as collators and carriers of signal information. Therefore, not all of the area of an imager is taken up by the photodiode. In fact, the percentage area of the detector that actually registers information within an individual pixel is referred to as “fill factor.” So although a pixel may have a nominal area, the fill factor may be of the order of 35%. Therefore, smaller pixels capture fewer x-ray photons and result in more image noise. Consequently, CBCT imaging using smaller matrix sizes usually requires greater radiation and higher patient dose exposure.

The resolution, and therefore detail, of CBCT imaging is determined by the individual volume elements or voxels produced from the volumetric data set. In CBCT imaging, voxel dimensions primarily depend on the pixel size on the area detector, unlike those in conventional CT, which depend on slice thickness. The resolution of the area detector is submillimeter (range: 0.09 mm to 0.4 mm), which principally determines the size of the voxels. Therefore, CBCT units, in general, provide voxel resolutions that are isotropic (equal in all three dimensions) (Fig. 3).

### *Image reconstruction*

Once the basis projection frames have been acquired, data must be processed to create the volumetric data set. This process is called reconstruction. The number of individual projection frames may be from 100 to



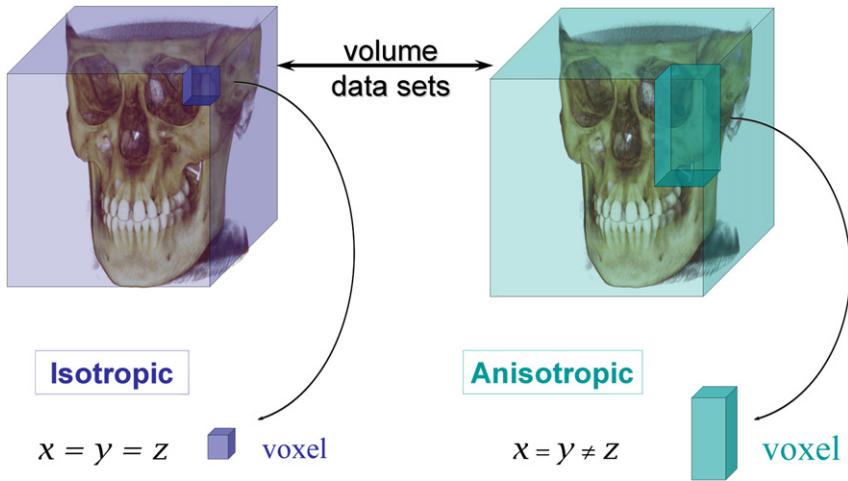


Fig. 3. Comparison of volume data sets obtained isotropically (*left*) and anisotropically (*right*). Because CBCT data acquisition depends on the pixel size of the area detector and not on the acquisition of groups of rows with sequential translational motion, the compositional voxels are equal in all three dimensions, rather than columnar with height being different from the width and depth dimensions.

more than 600, each with more than one million pixels, with 12 to 16 bits of data assigned to each pixel. The reconstruction of the data is therefore computationally complex. To facilitate data handling, data are usually acquired by one computer (acquisition computer) and transferred by way of an Ethernet connection to a processing computer (workstation). In contrast to conventional CT, cone-beam data reconstruction is performed by personal computer rather than workstation platforms.

Reconstruction times vary, depending on the acquisition parameters (voxel size, FOV, number of projections), hardware (processing speed, data throughput from acquisition to workstation computer), and software (reconstruction algorithms) used. Reconstruction should be accomplished in an acceptable time (less than 3 minutes for standard resolution scans) to complement patient flow.

The reconstruction process consists of two stages, each composed of numerous steps (Fig. 4).

#### *Acquisition stage*

Because of the spatially varying physical properties of the photodiodes and the switching elements in the flat panel, and also because of variations in the x-ray sensitivity of the scintillator layer, raw images from CBCT detectors show spatial variations of dark image offset and pixel gain. The dark image offset (ie, the detector output signal without any x-ray exposure), and its spatial variations are mainly caused by the varying dark current of the photodiodes. Gain variations are caused by the varying sensitivity of the photodiodes

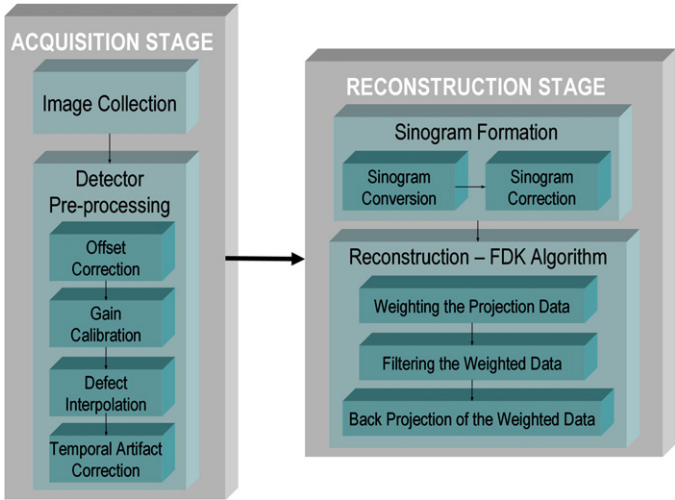


Fig. 4. Stages and steps involved in reconstruction of acquired projection data to form volumetric data from CBCT acquisition. The acquisition stage involves image collection and detector preprocessing, whereas the reconstruction stage involves sinogram formation and reconstruction using the FDK algorithm.

and by variations in the local conversion efficiency of the scintillator material caused by, for example, thickness or density variations. In addition to offset and gain variations, even high-quality detectors exhibit inherent pixel imperfections or a certain amount of defect pixels. To compensate for these inhomogeneities, raw images require systematic offset and gain calibration and a correction of defect pixels. The sequence of the required calibration steps is referred to as “detector preprocessing” (Fig. 5) and the calibration requires the acquisition of additional image sequences [8].

*Reconstruction stage*

Once images are corrected, they must be related to each other and assembled. One method involves constructing a sinogram: a composite image relating each row of each projection image (Fig. 6). The final step in the reconstruction stage is processing the corrected sinograms. A reconstruction filter algorithm is applied to the sinogram and converts it into a complete 2D CT slice. The most widely used filtered back projection algorithm for cone-beam-acquired volumetric data is the FDK algorithm [4]. Once all the slices have been reconstructed, they can be recombined into a single volume for visualization.

*Image display*

The availability of CBCT technology provides the dental clinician with a great choice of image display formats. The volumetric data set is

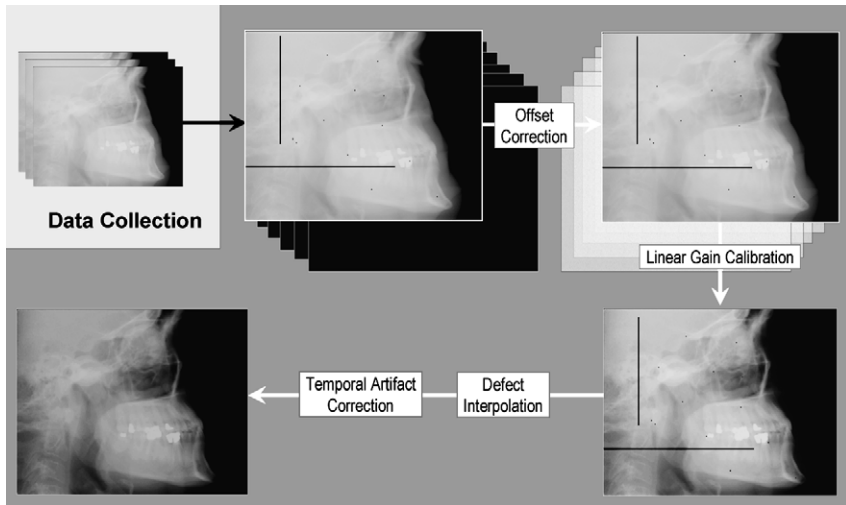


Fig. 5. CBCT detector preprocessing. The first step of the detector preprocessing is the offset correction, which is performed by pixel-wise subtraction of an individual offset value computed by averaging over a series of up to 30 dark images. The second step is the linear gain calibration, consisting of dividing each pixel by its individual gain factor. The gain factors are obtained by averaging a sequence, again with up to 30 images, of homogeneous exposures without any object between x-ray source and detector. The gain sequence is first offset corrected with its own sequence of dark images. The next procedure is the defect interpolation. Each pixel that shows unusual behavior, either in the gain image or in the average dark sequence, is marked in a defect map. The gray values of pixels classified as defective in this way are computed by linear interpolation along the least gradient descent. Flat detectors usually require an additional procedure to correct for temporal artifacts, which arise in flat detectors because the scintillator and photodiodes exhibit residual signals.

a compilation of all available voxels and, for most CBCT devices, it is presented to the clinician on screen as secondary reconstructed images in three orthogonal planes (axial, sagittal, and coronal), usually at a thickness defaulted to the native resolution (Fig. 7). Optimum visualization of orthogonal reconstructed images depends on the adjustment of window level and window width to favor bone and the application of specific filters.

### Advantages of cone-beam CT in dentistry

Being considerably smaller, CBCT equipment has a greatly reduced physical footprint and is approximately one quarter to one fifth the cost of conventional CT. CBCT provides images of highly contrasting structures and is therefore particularly well suited for the imaging of osseous structures of the craniofacial area. The use of CBCT technology in clinical dental practice provides a number of advantages for maxillofacial imaging.

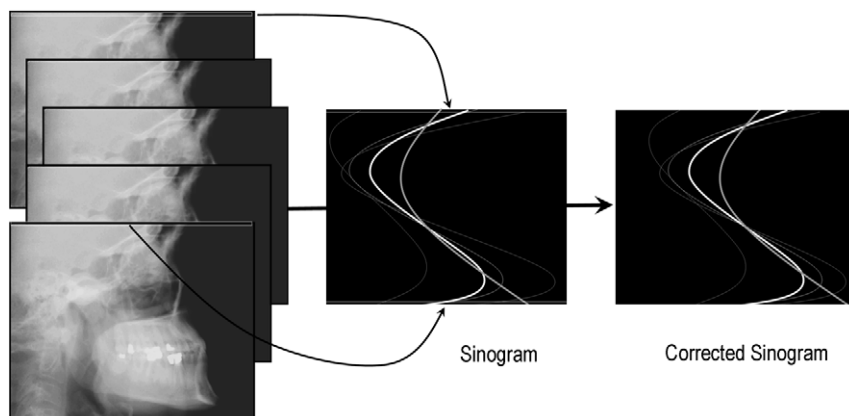


Fig. 6. Construction and correction of sinograms. This process is referred to as the radon transformation. Individual rows of each basis image are extracted and used to develop a composite image. The first row on the first basis image is used to form the first row on the composite image, the first row on the second image is used to form the second row on the composite image, and so on, until the first row on the  $n$ th image forms the  $n$ th row on the composite image. The resulting image comprises multiple sine waves of different amplitudes. The waves represent features in the object that are being rotated over  $360^\circ$ . However, before reconstruction, sinograms must be corrected for centering errors by detecting the skew in the sinogram and shifting it appropriately. Once corrected, the sinogram is ready to be processed by the appropriate reconstruction algorithm.

### *Rapid scan time*

Because CBCT acquires all projection images in a single rotation, scan time is comparable to panoramic radiography, which is desirable because artifact due to subject movement is reduced. Computer time for data set reconstruction, however, is substantially longer; it varies, depending on FOV, number of basis images acquired, resolution, and reconstruction algorithm, and may range from approximately 1 minute to 20 minutes.

### *Beam limitation*

Collimation of the CBCT primary x-ray beam enables limitation of the x-radiation to the area of interest. Therefore, an optimum FOV can be selected for each patient based on suspected disease presentation and region of interest. Although not available on all CBCT systems, this function is highly desirable because it provides dose savings by limiting the irradiated field to fit the FOV.

### *Image accuracy*

CBCT imaging produces images with submillimeter isotropic voxel resolution ranging from 0.4 mm to as low as 0.076 mm. Because of this

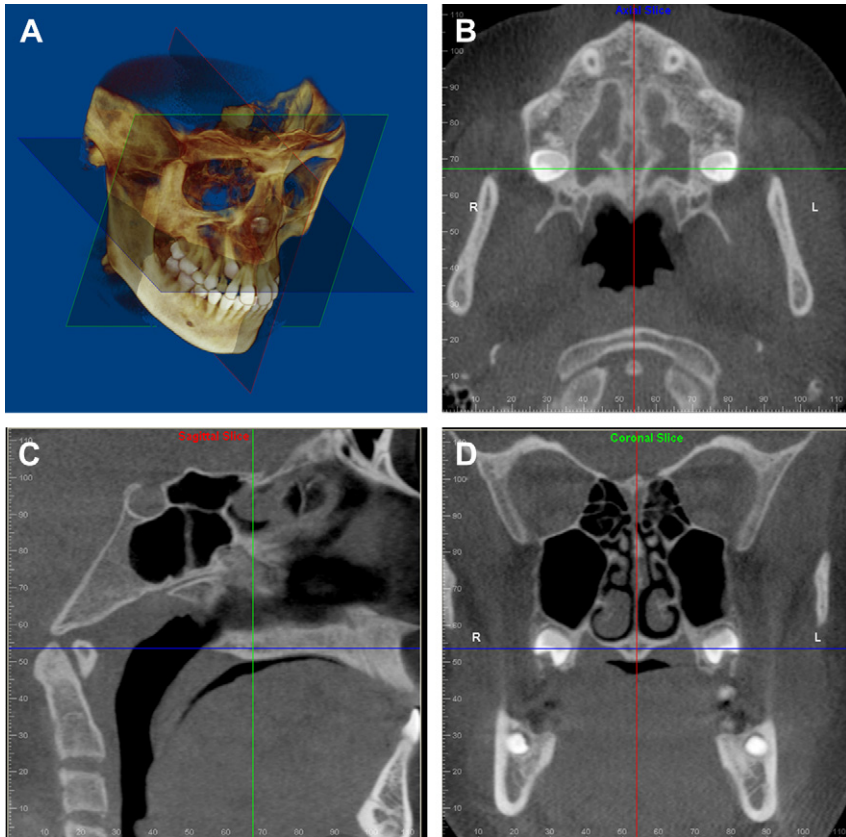


Fig. 7. Standard display modes of CBCT volumetric data. (A) Volumetric 3D representation of hard tissue showing the three orthogonal planes in relation to the reconstructed volumetric data set; each orthogonal plane has multiple thin-slice sections in each plane. (B) Representative axial image. (C) representative sagittal image, and (D) representative coronal image. (Images produced using Dolphin 3D, Chatsworth, California).

characteristic, subsequent secondary (axial, coronal, and sagittal) and multi-planar reformation (MPR) images achieve a level of spatial resolution accurate enough for measurement in maxillofacial applications where precision in all dimensions is important, such as implant site assessment and orthodontic analysis.

#### *Reduced patient radiation dose*

Published reports indicate that the effective dose [9] varies for various full FOV CBCT devices, ranging from 29 to 477  $\mu\text{Sv}$ , depending on the type and model of CBCT equipment and FOV selected (Table 2) [10–12]. Comparing these doses with multiples of a single panoramic dose or background equivalent radiation dose, CBCT provides an equivalent patient radiation dose of

Table 2  
Comparative radiation effective dose from selected cone-beam CT systems

CBCT unit	Technique	Dose <sup>a</sup>			
		Effective dose <sup>a</sup> (μSv)	Comparative		
			Imaging surveys	Annual per capita background <sup>c</sup>	
				No. of days	% Annual
CB MercuRay <sup>d</sup>	12-in/9-in/6-in FOV	477/289/169	74/45/26	48.0/29.0/17.0	13.0/8.0/4.7
Galileos <sup>e</sup>	Default/maximum	29/54	5/9	3.0/5.5	0.8/1.5
i-Cat <sup>d</sup>	12-in/9-in FOV	135/69	21/11	13.5/7.0	3.7/1.9
Iluma <sup>e</sup>	Low/high	61/331	10/53	6.2/33.5	1.7/9.2
Newtom 3G <sup>d</sup>	12-in/9-in FOV	45/37	7/6	4.5/3.5	1.2/1.0
PreXion 3D <sup>e</sup>	Standard/high-resolution	69/160	11/25	7.0/16.0	1.9/4.4
ProMax 3D <sup>e</sup>	Small/large	157/210	25/33	16.0/21.5	4.4/5.8

<sup>a</sup> Using 1990 International Commission on Radiological Protection calculations.

<sup>b</sup> *Data from* Ludlow JB, Davies-Ludlow LE, Brooks SL. Dosimetry of two extraoral direct digital imaging devices: NewTom cone beam CT and Orthophos Plus DS panoramic unit. *Dentomaxillofac Radiol* 2003;32:229–34.

<sup>c</sup> Annual per capita = 3.6 mSv (3600 μSv) per annum.

<sup>d</sup> *Data from* Ludlow JB, Davies-Ludlow LE, Brooks SL. Dosimetry of 3 CBCT devices for oral and maxillofacial radiology: CB Mercuray, NewTom 3G and i-CAT. *Dentomaxillofac Radiol* 2006;35:219–26.

<sup>e</sup> *Data from* Ludlow JB, Davies-Ludlow LE, Mol A. Dosimetry of recently introduced CBCT units for oral and maxillofacial radiology. *Proceedings of the 16th International Congress of Dentomaxillofacial Radiology*. Beijing, China, June 26–30, 2007. p. 97.

5 to 74 times that of a single film-based panoramic X ray, or 3 to 48 days of background radiation. Patient positioning modifications (tilting the chin) and use of additional personal protection (thyroid collar) can substantially reduce the dose by up to 40% [9,10]. Comparison with patient dose reported for maxillofacial imaging by conventional CT (approximately 2000  $\mu$ Sv) indicates that CBCT provides substantial dose reductions of between 98.5% and 76.2% [11–13].

### *Interactive display modes applicable to maxillofacial imaging*

Perhaps the most important advantage of CBCT is that it provides unique images demonstrating features in 3D that intraoral, panoramic, and cephalometric images cannot. CBCT units reconstruct the projection data to provide interrelational images in three orthogonal planes (axial, sagittal, and coronal). In addition, because reconstruction of CBCT data is performed natively using a personal computer, data can be reoriented so that the patient's anatomic features are realigned. Basic enhancements include zoom or magnification, window/level, and the ability to add annotation. Cursor-driven measurement algorithms provide the clinician with an interactive capability for real-time dimensional assessment. On-screen measurements provide dimensions free from distortion and magnification.

### *Multiplanar reformation*

Because of the isotropic nature of the volumetric data sets, they can be sectioned nonorthogonally. Most software provides for various nonaxial 2D images, referred to as MPR. Such MPR modes include oblique, curved planar reformation (providing “simulated” distortion-free panoramic images), and serial transplanar reformation (providing cross-sections), all of which can be used to highlight specific anatomic regions and diagnostic tasks (Fig. 8), which is important, given the complex structure of the maxillofacial region.

Because of the large number of component orthogonal images in each plane and the difficulty in relating adjacent structures, two methods have been developed to visualize adjacent voxels.

### *Ray sum or ray casting*

Any multiplanar image can be “thickened” by increasing the number of adjacent voxels included in the display, which creates an image slab that represents a specific volume of the patient, referred to as a ray sum. Full-thickness perpendicular ray sum images can be used to generate simulated projections such as lateral cephalometric images (Fig. 9). Unlike conventional X rays, these ray sum images are without magnification and are undistorted. However, this technique uses the entire volumetric data set, and interpretation suffers from the problems of “anatomic noise,” the superimposition of multiple structures.

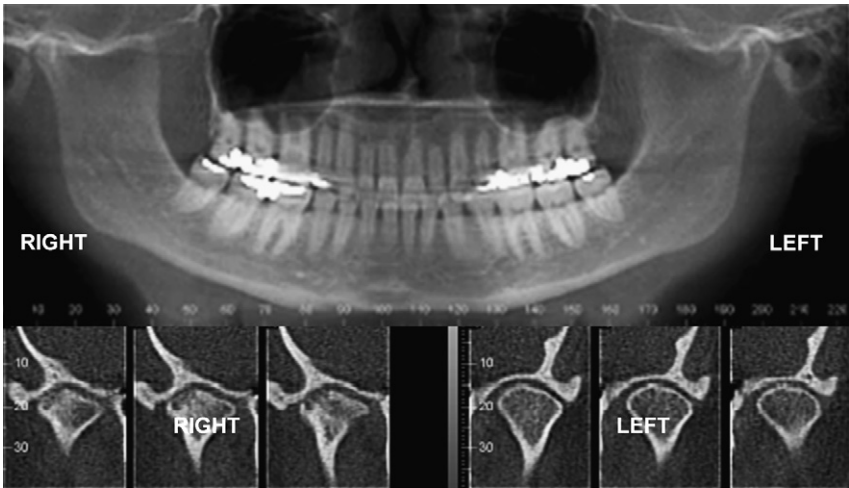


Fig. 8. Curved MPR simulated “panoramic” image from CBCT showing CBCT applications in temporomandibular joint assessment. Reformatted “panoramic” image (*top*) showing right side condyle differences in shape compared with normal left. Cropped paracoronal reformatted images clearly showing subcortical cystic defects in surface of right condyle as compared with the left, indicative of active degenerative joint disease. Images generated using i-CAT (Imaging Sciences International, Hatsfield, Pennsylvania).

*Three-dimensional volume rendering*

Volume rendering refers to techniques that allow the visualization of 3D data through integration of large volumes of adjacent voxels and selective display (Fig. 10). Two specific techniques are available.

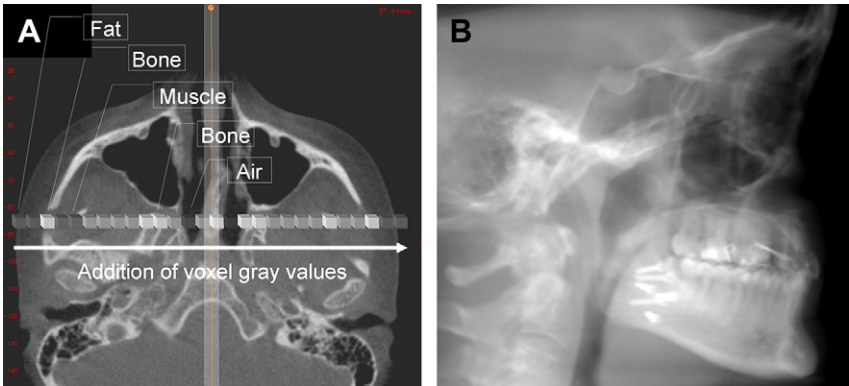


Fig. 9. Construction of ray sum images. An axial projection (*A*) is used as the reference image. A section slice is identified (*dashed line*) which, in this case, corresponds to the midsagittal plane and the thickness of this slice is increased to include the left and right sides of the volumetric data set. As the thickness of the “slab” increases, adjacent voxels representing elements such as air, bone, and soft tissues are added. The resultant image (*B*) generated from a full-thickness ray sum provides a simulated lateral cephalometric image.



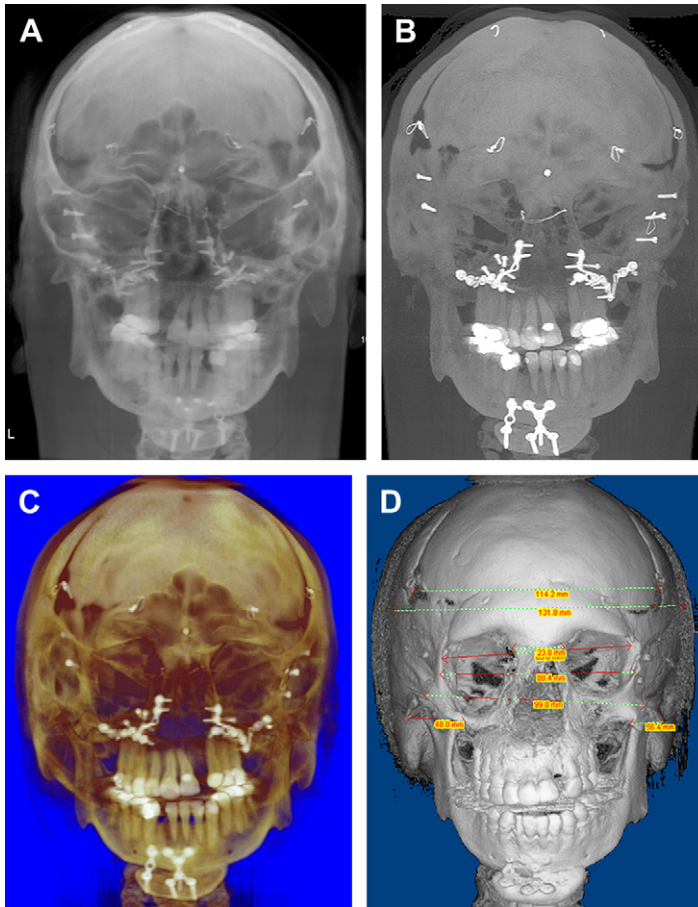


Fig. 10. Comparison of 3D visualization techniques applied to CBCT data set of a patient who has a craniofacial deformity. Anterior-posterior projection rendered using the direct volume rendering techniques of (A) ray sum and (B) maximum intensity projection and the indirect volume techniques of (C) volumetric transparent and (D) shaded surface display. Volumetric data set acquired with extended FOV i-CAT (Imaging Sciences International, Hatsfield, Pennsylvania) and all reconstructions generated using Dolphin 3D (Dolphin Imaging, Chatsworth, California).

**Indirect volume rendering.** Indirect volume rendering is a complex process, requiring selecting the intensity or density of the grayscale level of the voxels to be displayed within an entire data set (called segmentation). This technique is technically demanding and computationally difficult, requiring specific software; however, it provides a volumetric surface reconstruction with depth.

**Direct volume rendering.** Clinically and technically, direct volume rendering is a much more simple process. The most common direct volume rendering technique is maximum intensity projection (MIP). MIP visualizations are

achieved by evaluating each voxel value along an imaginary projection ray from the observer's eyes within a particular volume of interest and then representing only the highest value as the display value. Voxel intensities that are below an arbitrary threshold are eliminated.

### **Limitations of cone-beam CT imaging**

While clinical applications of CBCT have expanded, current CBCT technology has limitations related to the "cone-beam" projection geometry, detector sensitivity, and contrast resolution that produces images that lack the clarity and usefulness of conventional CT images. The clarity of CBCT images is affected by artifacts, noise, and poor soft tissue contrast.

#### *Artifacts*

An artifact is any distortion or error in the image that is unrelated to the subject being studied. Artifacts can be classified according to their cause.

#### *X-ray beam artifacts*

CT image artifacts arise from the inherent polychromatic nature of the projection x-ray beam that results in what is known as beam hardening (ie, its mean energy increases because lower energy photons are absorbed in preference to higher energy photons). This beam hardening results in two types of artifact: (1) distortion of metallic structures due to differential absorption, known as a cupping artifact, and (2) streaks and dark bands that can appear between two dense objects. Because the CBCT x-ray beam is heterochromatic and has lower mean kilovolt (peak) energy compared with conventional CT, this artifact is more pronounced on CBCT images. In clinical practice, it is advisable to reduce the FOV to avoid scanning regions susceptible to beam hardening (eg, metallic restorations, dental implants), which can be achieved by collimation, modification of patient positioning, or separation of the dental arches. More recently, dental CBCT manufacturers have introduced artifact reduction technique algorithms within the reconstruction process (eg, Scanora 3D, SOREDEX, Helsinki, Finland) (Fig. 11). These algorithms reduce image-, noise-, metal-, and motion-related artifacts and require fewer projection images, and therefore may allow for a lower acquisition dose. However, they are computationally demanding and require increased reconstruction times.

#### *Patient-related artifacts*

Patient motion can cause misregistration of data, which appears as unsharpness in the reconstructed image. This unsharpness can be minimized by using a head restraint and as short a scan time as possible. The presence of dental restorations in the FOV can lead to severe streaking artifacts. They

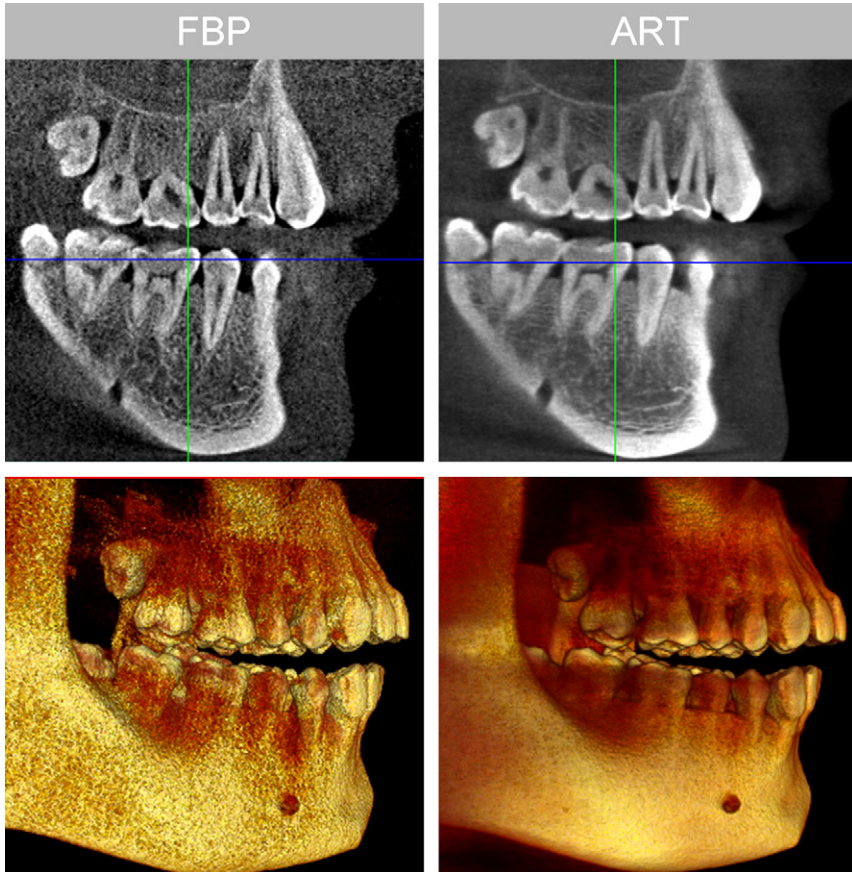


Fig. 11. Comparison of image quality of sagittal (*upper*) and 3D (*lower*) renderings reconstructed from 300 projection images using conventional Feldkamp back projection (FBP) (*left column*) and an iterative reconstruction called algebraic reconstruction technique (ART) (*right column*). While ART requires greater computing power, it also reduces artifacts requiring fewer projections to conduct the reconstruction (equals less dose) and is less sensitive to common patient movement and metal artifacts. (*Courtesy of SOREDEX, Helsinki, Finland; with permission.*)

occur because of extreme beam hardening or photon starvation due to insufficient photons reaching the detector, resulting in horizontal streaks in the image and noisy projection reconstructions. This problem can be reduced by removing metallic objects such as jewelry before scanning commences.

#### *Scanner-related artifacts*

Typically, scanner-related artifacts present as circular or ring-shaped, resulting from imperfections in scanner detection or poor calibration. Either of these two problems will result in a consistent and repetitive reading at each angular position of the detector, resulting in a circular artifact.

### *Cone beam-related artifacts*

The beam projection geometry of the CBCT and the image reconstruction method produce three types of cone-beam-related artifacts: (1) partial volume averaging, (2) undersampling, and (3) cone-beam effect.

*Partial volume averaging.* Partial volume averaging is a feature of conventional fan and CBCT imaging. It occurs when the selected voxel resolution of the scan is greater than the spatial or contrast resolution of the object to be imaged. In this case, the pixel is not representative of the tissue or boundary; however, it becomes a weighted average of the different CT values. Boundaries in the resultant image may present with a “step” appearance or homogeneity of pixel intensity levels. Partial volume averaging artifacts occur in regions where surfaces are rapidly changing in the z direction (eg, in the temporal bone). Selection of the smallest acquisition voxel can reduce the presence of these effects.

*Undersampling.* Undersampling can occur when too few basis projections are provided for the reconstruction. A reduced data sample leads to misregistration and sharp edges and noisier images because of aliasing, where fine striations appear in the image. This effect may not degrade the image severely; however, when resolution of fine detail is important, undersampling artifacts need to be avoided as far as possible by maintaining the number of basis projection images.

*Cone-beam effect.* The cone-beam effect is a potential source of artifacts, especially in the peripheral portions of the scan volume. Because of the divergence of the x-ray beam as it rotates around the patient in a horizontal plane, projection data are collected by each detector pixel. The amount of data corresponds to the total amount of recorded attenuation along a specific beam projection angle as the scanner completes an arc (Fig. 12). The total amount of information for peripheral structures is reduced because the outer row detector pixels record less attenuation, whereas more information is recorded for objects projected onto the more central detector pixels, which results in image distortion, streaking artifacts, and greater peripheral noise. This effect is minimized by manufacturers incorporating various forms of cone-beam reconstruction. Clinically, it can be reduced by positioning the region of interest adjacent to the horizontal plane of the x-ray beam and collimation of the beam to an appropriate FOV.

### *Image noise*

The cone-beam projection acquisition geometry results in a large volume being irradiated with every basis image projection. As a result, a large portion of the photons engage in interactions by way of attenuation. Most of these occur by Compton scattering producing scattered radiation.

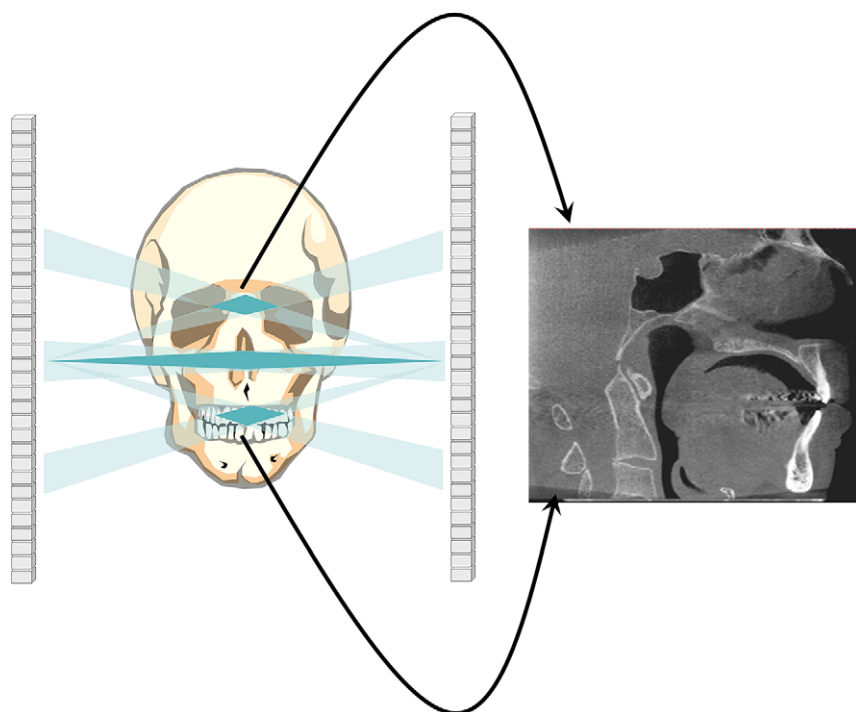


Fig. 12. Origin of the cone-beam effect. The projection of three x-ray beams (one perpendicular, one angled inferiorly, and the other angled superiorly) from a point origin are shown at two positions of the x-ray tube, 180° apart. The amount of data collected by the detector for reconstruction corresponds to the solid volume between the overlapping projections. Centrally, the amount of data acquired is maximal, whereas peripherally, the amount of data collected is appreciably less. The midsagittal section image demonstrates the visual effects of this in producing a peripheral “V” artifact of increased noise, distortion, and reduced contrast.

Most of the scattered radiation is produced omnidirectionally and is recorded by pixels on the cone-beam area detector, which does not reflect the actual attenuation of the object within a specific path of the x-ray beam. This additional recorded x-ray attenuation, reflecting nonlinear attenuation, is called noise. Because of the use of an area detector, much of this nonlinear attenuation is recorded and contributes to image degradation or noise. The scatter-to-primary ratios are about 0.01 for single-ray CT and 0.05 to 0.15 for fan-beam and spiral CT, and may be as large as 0.4 to 2.0 in CBCT. Problems also exist with detectors and algorithms.

#### *Poor soft tissue contrast*

Three factors limit the contrast resolution of CBCT. Although scattered radiation contributes to increased image noise, it is also a significant factor in reducing the contrast of the cone-beam system. In addition, the divergence of the x-ray beam over the area detector causes a pronounced heel

effect. This effect produces a large variation in, or nonuniformity of, the incident x-ray beam on the patient and resultant nonuniformity in absorption, with greater signal-to-noise ratio (noise) on the cathode side of the image relative to the anode side. Finally, numerous inherent flat-panel detector-based artifacts affect its linearity or response to x-radiation. Although these conditions limit the application of current CBCT imaging to the assessment of osseous structures, several techniques and devices are currently being investigated to suppress this effect.

## **Applications**

Currently, CBCT is used most commonly in the assessment of bony and dental pathologic conditions, including fracture; structural maxillofacial deformity and fracture recognition; preoperative assessment of impacted teeth; and temporomandibular joint imaging; and in the analysis of available bone for implant placement. In orthodontics, CBCT imaging is now being directed toward 3D cephalometry. These applications are described in detail in other articles.

The availability of CBCT is also expanding the use of additional diagnostic and treatment software applications, all directed toward 3D visualization, because CBCT data can be exported in the nonproprietary Digital Imaging and Communications in Medicine file format standard. CBCT permits more than diagnosis; it facilitates image-guided surgery. Diagnostic and planning software are available to assist in orthodontic assessment and analysis (eg, Dolphin 3D, Dolphin Imaging, Chatsworth, California) and in implant planning to fabricate surgical models (eg, Biomedical Modeling Inc., Boston, Massachusetts); to facilitate virtual implant placement; to create diagnostic and surgical implant guidance stents (eg, Virtual Implant Placement, Implant Logic Systems, Cedarhurst, New York; Simplant, Materialise, Leuven, Belgium; EasyGuide, Keystone Dental, Burlington, Massachusetts); and even to assist in the computer-aided design and manufacture of implant prosthetics (NobelGuide/Procera software, Nobel Care AG, Göteborg, Sweden). Software is also available to provide surgical simulations for osteotomies and distraction osteogenesis (Maxilim, Medicim NV, Mechelen, Belgium). This area is a blossoming field that provides opportunities for practitioners to combine CBCT diagnosis and 3D simulations with virtual surgery and computer-assisted design and manufacture. Image guidance is an exciting advance that will undoubtedly have a substantial impact on dentistry.

## **Clinical implications**

The impact that CBCT technology has had on maxillofacial imaging since its introduction cannot be underestimated, which does not imply that CBCT is appropriate as an imaging modality of first choice in dental practice. However, no specific patient selection criteria have been published



for the use of CBCT in maxillofacial imaging (ie, guidelines as to when, where, why, what, how, and on whom). Because cone-beam exposure provides a radiation dose to the patient higher than any other imaging procedure in dentistry, it is paramount that practitioners abide by the ALARA principle. The basis of this principle is that the justification of the exposure to the patient is that the total potential diagnostic benefits are greater than the individual detriment radiation exposure might cause. CBCT should not be considered a replacement for standard digital radiographic applications that, ironically, also use a cone beam of radiation, but without computed integration of basis projections. Rather, CBCT is a complementary modality for specific applications.

Although deceptively simple, the technical component of patient exposure is only one half of cone-beam imaging. Based on the medical model of imaging, a moral, ethical, and legal responsibility of interpreting the resultant volumetric data set exists [14–16]. The mechanics of interpretation involve image reporting with the development of a series of images formatted to display the condition/region appropriately (image report) and a cognitive interpretation of the significance of the imaging findings (interpretive report). These skills are not within the domain of most general and specialist practitioners; however, they act as the de facto standard of care in providing CBCT services. It would behoove those contemplating or currently using CBCT imaging to develop and maintain these skills.

## Summary

The development and rapid commercialization of CBCT technology dedicated for use in the maxillofacial region will undoubtedly increase general and specialist practitioner access to this imaging modality. CBCT is capable of providing accurate, submillimeter-resolution images in formats allowing 3D visualization of the complexity of the maxillofacial region. All current generations of CBCT systems provide useful diagnostic images. Future enhancements will most likely be directed toward reducing scan time; providing multimodal imaging (conventional panoramic and cephalometric, in addition to CBCT images); improving image fidelity, including soft tissue contrast; and incorporating task-specific protocols to minimize patient dose (eg, high-resolution, small FOV for dentoalveolar imaging or medium-resolution, large FOV for dentofacial orthopedic imaging). The increasing availability of this technology provides the practitioner with a modality that is extending maxillofacial imaging from diagnosis to image guidance of operative and surgical procedures.

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