Basics of CBCT Imaging

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Abstract

Cone Beam Computed Tomography (CBCT) is now widely available to dentists for examining hard tissues in the dental and maxillofacial regions. CBCT gives a three-dimensional view of anatomy and pathology. It uses a cone shaped source of ionizing radiation and a two-dimensional area detector fixed on a rotating gantry to acquire multiple sequential projection images in one complete scan around the area of interest. It is important, as an users to understand the basic concepts of this imaging modality. This review covers the basics of dental CBCT imaging including the hardware used, image acquisition,image reconstruction and image visualization.

Key Words: CBCT, Gantry, Flat panel detectors, FDK algorithm, Multiplanar reformatting.

Introduction

Cone Beam Computed Tomography (CBCT) is a recent technology, which was first applied for angiography in the early 1980s and then later gradually used for other applications.[1] CBCT allows the three dimensional assessment of hard tissues of the maxillofacial region.[2] The introduction of CBCT imaging has heralded a shift from a two-dimensional to a volumetric approach in maxillofacial imaging.[3] CBCT has been available in maxillofacial radiology for more than a decade. Ample models of equipment are in existence, and there is evidence of widespread use in some countries. Compared to conventional dental radiography in dentistry, CBCT imaging has higher radiation doses and it becomes even more important to adhere to the radiation protection principle of justification.[4] Its widespread use has raised the concerns regarding justification and optimization of CBCT exposures, training of the users and quality assurance of the scanners. Therefore, it is important to have at least the basic understanding of the technical aspects of dental CBCT imaging in order to reap the full benefit of this technique while minimizing radiation related risk.[1]

This review offers an outline of technical aspects of dental CBCT imaging including hardware, acquisition, image reconstruction and visualization.

Hardware:

X-ray Generation (X-ray tube)

An x-ray tube is composed of a cathode and an anode placed in a vacuum glass tube. Cathode consists of tungsten filament of 2mm in diameter and 1 cm in length, which lies in a molybdenum focusing cup. The anode consists of tungsten target, which is embedded in copper stem (Fig. 1).[4] The filament is heated to incandescence by application of low voltage tube current of about 10 volts, which emits electrons at a rate proportional to the temperature of the filament.

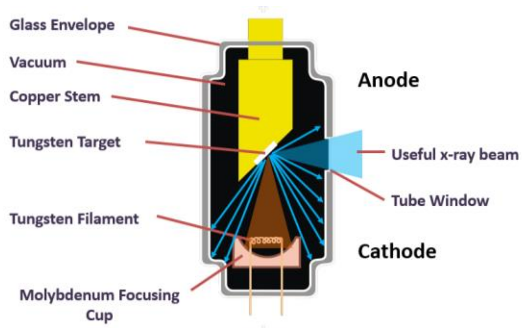

Fig.1 X-Ray Tube

A high voltage of 90 to 120 kV is applied between cathode and anode causing the electrons to move. These electrons are electrostatically focused by the molybdenum cup into a narrow beam and are directed to a small rectangular area on the anode called the focal spot. Most of these electrons traveling from cathode to anode interact with target electrons and release their energy as heat. Only a small number of these electrons convert their kinetic energy into X-ray photons by the formation of bremsstrahlung radiation and characteristic radiation.[3]

To limit the exposure of patients' to radiation, the X-ray beam is collimated by blocking all those that are not passing through the scanned volume. This is done using a lead collimator that has an opening for X-rays to pass through. Most CBCT systems have multiple pre-defined field-of-view (FOV) sizes and hence a collimator will have several pre-defined openings according to the FOV sizes.[1]

Gantry (Fig. 2)

Most dental CBCT machines use a setup in which X-ray tube and detector are connected in the horizontal plane, allowing for seated and/or standing patient positioning (fixed C-arm).[1]Based on the type of the unit scans are made with the patient in supine, sitting, or standing position. Supine units are physically larger in size and may not be possible to accommodate physical disabled patients. Units with seated type are more comfortable but still they may not allow scanning of wheelchair seated or physically disabled patients. On the other hand standing units may not be able to be adjusted low enough to accommodate wheelchair seated patients

Fig.2 CBCT **machines – standing type of unit**

. In any setup immobilizing the patient's head is more important than patient positioning as any movement during scan degrades the final image. Immobilization of the head is achieved by using a chin cup, bite fork, or other head-restraint mechanism.[3]

To station the FOV according to the region of interest (ROI), limited amount of translator movement of the C-arm is usually possible within this plane as well as up–down movement, especially for scanners with a small FOV.[1] The dimensions of scan volume or FOV 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 cylindrical or spherical.[5]

CBCT systems can be categorized according to the available FOV or selected scan volume height (Fig 3) as follows:[5]

Localized region: approximately ≤ 5 cm (e.g.

dentoalveolar, temporomandibular joint)

Single arch: 5 - 7 cm (e.g. maxilla/ mandible)

Interarch: 7 - 10 cm (e.g. mandible and superiorly to include the inferior concha)

Maxillofacial: 10 - 15 cm (e.g. mandible and extending to Nasion)

Craniofacial: > 15 cm (e.g. from the lower border of the mandible to the vertex of the head)

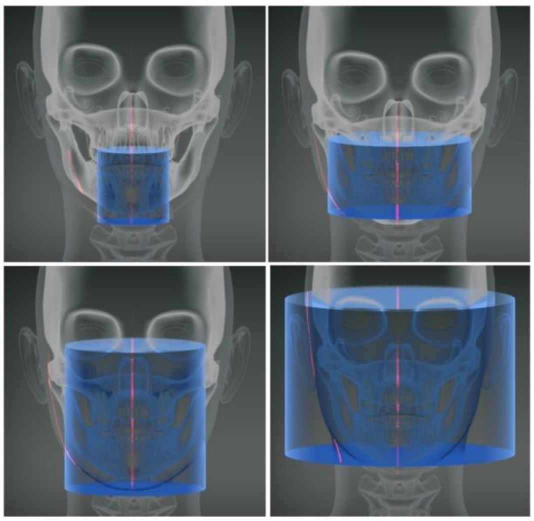

Fig. 3 Various Field of Views used in CBCT scan

Detector

X-ray detectors convert the incoming X-ray photons into an electrical signal. Current CBCT machines use either of the following detector type: (1) image intensifier tube/charge-coupled device (IIT/CCD) combination or (2) flat panel detectors (FPDs).[3]

The IIT/CCD configuration comprises an x-ray IIT coupled to a CCD by way of a fiber optic coupling.[5] The input phosphor screen converts X-ray beam into an optical signal which is then converted to electrons by the photocathode screen. Electrons are accelerated through the electric field inside the image intensifier and get converted to an optical signal at the output phosphor screen. The optical iris adjusts the optical signal which is detected by the CCD. Geometrical distortion and a blurring component of veiling glare generated at the image intensifier is the final component of the read-out image.[6]

The flat-panel detector technology used in CBCT was first investigated by Jaffray and Siewerdsen in 2002. This technology is based on fabricating 2D matrix of hydrogenated amorphous silicon thin-film transistors (TFTs) on a large area of scintillating material (Thallium doped Cesium Iodide). Such a setup demonstrate excellent efficiency of converting light photons into electrical signals and in readout signal (optical coupling efficiency) and hence improved imaging is possible with high uniformity over large area, high optical absorption, and high detective quantum efficiency (DQE) of approximately 60%.[7] More recently, large complementary metal oxide semiconductor (CMOS) technology arrays have also been used.[3] FPDs are distortion free, have a higher dose efficiency, a wider dynamic range and can be produced with either a smaller or larger FOV.[1] The FPD CBCT system has a higher spatial resolution than the IIT/CCD.[6]

Different components and technologies can be used to read-out the signal in FPDs, and a distinction can be made between CCD, TFT, CMOS and FPDs. These technologies differ in terms of pixel size, detector size, sensitivity, noise level and read-out speed and have varying cost efficiency depending on the total size of the detector. CCDs offer high-speed read-out at high resolution, but they are limited to a small FOV, and expanding the FOV tends to reduce dose efficiency. FPDs based on active matrix TFT read-out have also been incorporated in many applications of CBCT. More recently, CMOS detectors with a large FOV, high-speed read-out, fine resolution and low electronic noise are becoming available.[1]

Voxel Size: The individual volume elements (voxels) produced in formatting the volumetric data set determines the spatial resolution. CBCT units in general provide voxel resolutions that are isotropic i.e. equal in all three dimensions.[3]

Image Acquisition

CBCT imaging is based on acquiring the

projection data where the X-ray source and detector are attached on a C shaped gantry capable of performing a motorized movement around the patient. During the exposure, following a circular path covering an angular range of at least 200°, multiple x-ray projection images of the object are acquired along different angular directions.[8] The acquired images appear similar to lateral and Posterio-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 constituting the projection data throughout the scan is determined by the detector frame rate, the completeness of the trajectory arc, and the rotation speed of the source and detector.[3]

Exposure: Several CBCT devices use pulsed exposure, resulting in a large discrepancy between scan time (time between the first and last projection) and exposure time (the cumulative time during which an exposure is made). Other X-ray tubes allow only continuous exposure, for which the total scan time and exposure time are same. Both pulse and continuous exposure approaches are susceptible to effects of detector lag, but pulsed X-ray systems may exhibit improved spatial resolution owing to reduced motion effect (i.e. motion of the gantry during each exposure/read-out frame).

Frame rate and speed of rotation: Adjusting the detector frame rate to increase the number of basis image projections results in reconstructed images with fewer artifacts and better image quality. Images with less noise and reduced metallic artifacts are obtained because of higher frame rates (increased signal-to-noise ratio). But the drawback of higher frame rate is, it requires a longer scan time which increases the exposure dose to the patient. In addition, more data obtained, increases the primary reconstruction time.[3] Since normally, arc of exposure and exposure time is short, it is essential that detector pixels must be sensitive enough to capture adequate radiation to register a high signal-to-noise output and to transmit the voltage to the analog and the digital converter. In a clinical setting, the total number of available view angles is normally limited to several hundred due to the limitations of solid-state detector readout speed and the need for short scanning time.[5]

Completeness of the trajectory arc: Most CBCT machines use a complete 360° trajectory to acquire adequate projection data. Many CBCT have scan arcs less than 360^0. Most CBCT units have fixed scan arcs; some may provide a choice of manual controls to reduce the scan arc. A limited scan arc potentially reduces the scan time and patient radiation dose and is mechanically easier to perform.[3] However, in terms of image quality, a partial rotation tends to decrease overall image quality. Depending on the mA, a 180° rotation protocol can lead to a slight or more pronounced increase in noise than in a 360° protocol.[1]

It is desirable to reduce CBCT scan times to as short as possible to reduce motion artifact resulting from patient movement. Decreased scanning times may be achieved by increasing the detector frame rate, reducing the number of projections, or reducing the scan arc.[3]

Dose and Exposure settings: though the risk from dentomaxillofacial imaging is considered small for an individual, but when you multiply it by the large population of patients who are exposed to diagnostic imaging, radiation risk becomes a significant public health issue.[9] The radiation doses from dental CBCT are generally higher than in conventional dental radiography (such as intraoral, panoramic, cephalometric radiography) but lower than in Multiple detector computed tomography of the dental area. The dose is dependent on equipment type and exposure settings, especially the FOV, exposure time (s), tube current (mA) and the energy/potential (kV).[10]

Although tube current may be increased in some units and is suggested to compensate for increases in patient size, it leads to increase in the effective dose proportionately. Adjustment of kVp has an even greater effect on dose than mA, with each increase in 5 kVp approximately doubling the dose if all other parameters remain the same.[3] Dosimetry for standard CBCT exposure settings demonstrated significant reductions in effective dose associated with the use of small FOV sizes.[9] At the present only a single diagnostic reference level of 250 mGy cm^2 for the placement of an upper first molar implant in adults is available.[10] Exposure parameters should be appropriate for both the given patient size and the diagnostic task that motivated image selection.

Image Reconstruction

Before reconstruction, the acquired raw data or 2D projection data can undergo several pre-processing steps. The steps of preprocessing may vary between manufacturers. These steps are typically performed to remove aberrations associated with variations in detector pixel defects, gain and dark current.[1] In cone-beam geometry, 3D volumetric data can be directly reconstructed from the 2D projection data. This is referred to as cone-beam reconstruction. The most popular approximate reconstruction technique for cone-beam projections about a fixed isocenter acquired along a circular trajectory is the Feldkamp, Davis and Kress (FDK) algorithm. In this method, the measured cone-beam projections are pre-weighted, filtered and finally back projected along the same ray geometry as initially used for forward projection.[7]

The reconstruction process consists of two stages, each comprising numerous steps:[3]

1) **Preprocessing stage -** After the multiple planar projection images are acquired, these images must be corrected for inherent pixel imperfections, variations in sensitivity across the detector, and uneven exposure.

2) **Reconstruction stage -** The corrected images are converted into a special representation called a sinogram, a composite image developed from multiple projection images. The horizontal axis of a sinogram represents individual rays at the detector, whereas the vertical axis represents projection angles. If there are 300 projections, the sinogram will have 300 rows. This process of generating a sinogram is referred to as the Radon transformation. The resulting image comprises multiple sine waves of different amplitude, as individual objects are projected onto the detector at continuously varying angles. The final image is reconstructed from the sinogram with a filtered back-projection algorithm for volumetric data acquired by CBCT imaging; the most widely used algorithm is the Feldkamp algorithm. This process is referred to as inverse Radon transformation. When all slices have been reconstructed, they are combined into a single volume for visualization.

Image Visualization

Multiplanar reformatting

After reconstruction process a 3D matrix is created that can be viewed as a series of 2D cross-sectional images – axial (slices from top to bottom), sagittal (left to right) and coronal (anterior to posterior) views (fig 4).[1]

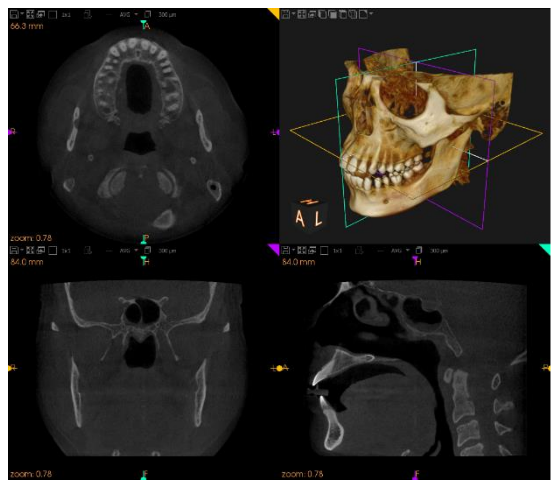

Fig. 4 Orthogonal sections

The volumetric data sets can be sectioned non-orthogonally since it is isotropic in nature. Most software provides for various nonaxial 2D images, referred to as Multi Planar Reformation (MPR). Such MPR modes include oblique, curved planar reformation and serial transplanar reformation (providing cross-sections).[5]

Oblique planar reformation creates nonaxial 2D images by transecting a set or "stack" of axial images.

This mode is particularly useful for evaluating specific structures (e.g. TMJ, impacted third molars, winding angles of the mandibular canal) (fig 5).[11]

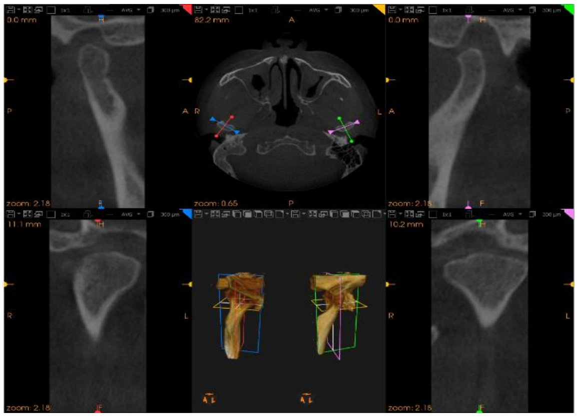

Fig.5 Oblique planar reformation to view TMJ

In order to generate a curved planar images, on an appropriate axial image we can manually draw a planning line by selecting multiple nodes along the centerline corresponding to the jaw arch; this creates a “simulated” or reconstructed dental panoramic image (fig 6).[3]

Fig. 6 Curved planar image

Serial trans-planar reformation produces a series of stacked sequential cross-sectional images orthogonal to the oblique or curved planar reformation.[11]

Since the number of component orthogonal images in each plane is large and there is difficulty in relating adjacent structures, two methods have been developed to visualize adjacent voxels.[5]

1) Ray sum or ray casting - by increasing the number of adjacent voxels included in the display of any multiplanar image it gets ‘‘thickened’’. This process creates an image slab that represents a specific volume of the patient and is referred to as a ray sum. By using full thickness perpendicular ray sum images we can generate simulated projections such as lateral cephalometric images

2) Three-dimensional volume rendering*:* refers to the techniques that allow the visualization of 3 dimensional data through integration of large volumes of adjacent voxels and selective display. Two specific techniques are available. (a) Indirect volume rendering - a complex process, requiring selecting the intensity or density of the grayscale level of the voxels to be displayed within an entire data set and provides a volumetric surface reconstruction with depth. (a) Direct volume rendering - most common direct volume rendering technique is maximum intensity projection (MIP). MIP visualizations are accomplished 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.

Conclusion

The development and rapid commercialization of CBCT technology has increased dentists access to this imaging modality. CBCT provides accurate, high resolution images in formats which permit 3 dimensional display of the complex anatomy of maxillofacial region. As the use of CBCT imaging becomes more of a routine procedure rather than a rarity, it necessitates the need to have basic knowledge to interpret these images. Hence this review outlined the basic concepts of CBCT imaging. Although the basic working mechanism will remain the same, future enhancements will most likely focus on reducing scan time, reducing patient exposure dose, improving various image quality aspects.

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