7 Basic Principles of Computer Tomography (MDCT/CBCT). The Use of MDCT and CBCT in Dentomaxillofacial Imaging

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7.1 Conventional Tomography

Conventional tomography is an imaging modality in which the production of the radiographic image is achieved by the combined movement of an X-ray tube and an X-ray receptor around an imaginary axis or point that constitutes the tomographic center. In the frame of the tomograph, the X-ray tube is joined by a fixed arm to the X-ray receiver so that a smooth, simultaneous movement between them is achieved in the opposing directions. The space between the X-ray tube and the X-ray receptor constitutes the tomographic field [1], within which the scanner produces sections of the anatomical structures of interest. The generation of the various sections is achieved by intentionally blurring all structures outside the tomographic field, while in contrast, structures within the tomographic field are projected sharply at a fixed position on the X-ray receptor. A prerequisite for avoiding distortions in the resulting tomographic images is the positioning of the anatomical structures under examination as perpendicular as possible to the source of the rays [2].

The major factors that affect the quality of the resulting tomographic images are the type of movement of the X-ray tube, the thickness of the sections, and the degree of magnification [3].

The X-ray tube can move in a linear, circular, ellipsoidal, sub-cyclical or spiral trajectory. With the exception of the linear type of movement in which the X-ray tube moves in a horizontal or vertical straight line, in all other types of movements mentioned beforetake place in more than one plane. Complex-nonlinear motion types produce more uniform ambiguity in the tomographic image [3–5]. An advantage of the complex motion types mentioned before is that images are produced without unwanted projections and shadows [6].

The thickness of the tomographic slice produced depends on the tomographic angle. Wide angles of 50–60° produce thin tomographic sections of 1–2 mm, whereas as the velocity of the X-ray tube decreases the radiographic angle is blunted and the sections increase in thickness.

Finally, the degree of magnification in conventional tomography varies, depending on the scanner, from 10% to 30% and is constant and comparable for all anatomical structures within the same scanning field [7].

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Conventional tomography has the disadvantages of increased patient exposure to radiation, magnification of the image, reduced resolution of details in the areas of similar density due to overlap and radiolucencies, and lack of features in each generated section to facilitate identification of the acquisition area [8].

7.2 Computed Tomography

7.2.1 Principles of Operation and Parts of the Tomograph

Computed tomography or CT is a digital imaging modality, which was first applied for brain tumor diagnosis in 1971 in the UK [9]. The main creators of computed tomography, Hounsfield and Cormack, were awarded the Nobel Prize in 1979.

A computed or CT tomographic scanner reconstructs images from various projections obtained using a rotating X-ray beam and software that uses complex mathematical algorithms. This enables it to produce sections of varying thicknesses of the anatomical structure under examination without the typical lack of definition and the overlap effect caused by neighboring anatomical structures of similar or higher density found in conventional tomography.

All CT scanners consist of the scanning system, the computing system, and the image display and recording system.

The scanning system of the CT scanner includes an X-ray tube, the radiation detector array and the examination table. The X-ray tube emits a confined fan-shaped X-ray beam while rotating 360° running on a ring approximately 1.5 m in diameter. The radiation detectors are attached to the same ring in an opposite position to the tube and collect the attenuated X-rays as they exit the structures of interest. The radiation detectors are mounted in rows and are usually either fluorescence or ionization chamber type. The movement of the X-ray tube is usually helical or spiral and continues until the area of interest is covered without interrupting the irradiation as was the case with older scanners where irradiation was interrupted and the examination table moved [4]. Finally, the examination table is constructed so that it can be moved along its axis and is positioned perpendicular to the plane defined by the tube and the radiation detectors.

The calculation system measures the attenuation of the X-ray beam after its passing through the structures under examination. The attenuation of the beam is not uniform but depends on the composition and density of the tissues it passes through. The computer of the tomograph receives the attenuation values from the array of detectors and then converts them into images. To create the image, the computer forms a matrix composed of a large number of pixels (elementary voxels). Each matrix is characterized by its size, i.e. by a certain number of pixels in each row and column. The computer first records the various linear attenuation coefficients of the X-rays as they exit the structures of interest. The linear attenuation coefficient is shown in the following formula, often referred to as Beer–Lambert's law.

$$I = I_0 e - \mu\chi$$

where I represents the intensity of the rays as they exit the structure of interest, I_0 is the intensity of the rays as they enter the structure of interest, e is Euler's constant equal to 2.718, μ is the linear attenuation coefficient or absorption coefficient, and χ is the thickness of the structure through which the rays pass. The resulting values of the linear coefficient are corresponded by the computer to grey color tones according to the Hounsfield scale and are called Hounsfield units (HU). Hounsfield units usually take values from −1000 which translates to zero attenuation of the rays to +1000 which translates to full absorption of the radiation [5]. In general, structures that exhibit high absorption of rays correspond to bright regions of the scale, while conversely structures that do not absorb rays to a large extent correspond to dark regions. The sum of Hounsfield's numbers from white to black constitute the window range which affects the contrast of the imaging. In reference, water has been defined as a value of 0, air −1000, adipose tissue from −50 to −100, parenchymal organs from +40 to +80, spongy bone from +300 to +800, and cortical bone values greater than +800.

These resulting values are included as tones of grey now within each of the matrix's generated volume cells.

The recording and display system, once the area of interest and the section have been selected, converts the 3D matrix into a two-dimensional representation on the computer screen. In the resulting image, the pixels of the selected section are now presented as small squares called pixels. Each pixel corresponds to the cross-sectional surface of the cross-sectional area of the matrix [5].

An important feature of computed tomography is that the original cross-sections taken during the movement of the tube are stored and can be reconstructed by the computer of the scanner in a different plane and direction (multi-planar reformatting), thus enabling the imaging of the structures of interest in different sections depending on the diagnostic need. Although the images resulting from multi-planar reformatting do not have the quality of the original CT images they still have a high diagnostic value.

7.2.2 Image Characteristics

The image quality in any tomograph depends on the spatial resolution capability, contrast resolution capability, noise, and blurring [10].

Spatial resolution is the ability of the scanner to distinguish small objects or structures that are very close to each other and is measured by line pairs per millimeter (l/mm). The spatial resolution is inversely proportional to the thickness of the section.

Contrast resolution is the ability of the scanner to distinguish objects or structures with very small differences in density. Computed tomography scanners, because of their large window range (−1000 to +1000), have the ability to distinguish anatomical structures with a difference in density of <1% [5].

Noise is defined as all information contained in a tomographic image that has no diagnostic value, while the sum of diagnostically useful information in an image is called signal. The sharpness of a tomographic image also depends on the signal-to-noise ratio (SNR). Noise cannot be completely eliminated, but with the evolution of scanners and their software it can be greatly reduced. So by reducing the noise or increasing the signal we can increase the quality of the tomographic image.

The lack of sharpness of the tomographic image can be the result of various factors such as: patient movements (physical or breathing movements), non-cooperation of the patient during the examination, the thickness of the anatomical structures under examination, the thickness of the slice chosen, limitations of the electronic parts (detection system, etc.) and the software of the tomograph, as well as the active size of the focus of the X-ray tube.

7.2.3 Evolution of Computed Tomography

Computed tomography scanners have gone through several stages of evolution and have incorporated various technologies and improvements in both their electronic components and software.

The first-generation scanners had a radiation detector and used a thin well-focused pencil-like X-ray beam. The scanning of the structure of interest was done by successive projections, with the tube-detector system rotating by 1° to create each of them. To create a section, 180 consecutive projections were required, resulting in an average examination time of 25–30 min.

In the second-generation scanners, the number of detectors was increased to 30 in an arc arrangement and the shape of the X-rays emitted by the tube took the form of a fan, which made it possible to reduce the average examination time to less than 90 s.

In the third generation, the number of detectors initially reached 288, while in more sophisticated scanners it rose to 700 in an arc configuration. The range of the beams was increased and adapted to cover the entire object under examination and thus, together with the increase in the number of detectors, a further reduction in the average examination time to about 5 s was achieved.

In the fourth generation, which consists of the most modern CT scanners, the number of radiation detectors has increased to over 2000 and the detectors are now mounted on an outer ring and do not move during the examination allowing to overcome the problems caused to the image by the movement of the scanning system. The average examination time was reduced to a few seconds.

In the third and fourth generation CT scanners, various improvements have been made, such as the spiral Ct scan, whereby the patient is moved as the tube-detector system rotates. The collection of information from the detectors is continuous and is not interrupted during the examination and thus the examination time is significantly reduced resulting in an improved image due to the reduction of distortions caused by the patient's physical and respiratory movements [11, 12]. Another improvement introduced in modern scanners is dual energy computed tomography (Dect), in which depending on the scanner, single or dual beam sources are used, as well as detectors with the ability to detect data of different energies. The advantage of dual energy computed tomography is the reduction of the noise, increase the ability to characterize and distinguish tissues, reduce artefacts resulting from the hardening of the beams, and further reduce the dose of beams received by the patient without negatively affecting the quality of the images [13].

7.2.4 Applications of Computed Tomography in Dental Science

In dental science, computed tomography has found applications in the identification of lesions of the jaws and adjacent anatomical structures, the assessment of occlusion of teeth, the diagnosis of temporomandibular joint disorders, the preoperative study for the placement of implants and in the postoperative assessment of the health of peri-implant bone structures [14]. Also, due to the zero magnification and the ability to match each section with the acquisition area, it enables the measurement of height and thickness of the bone, the estimation of its density and being able to transfer and match the information to the bony structures of interest on the patient. Finally, the ability of tomographs to reconstruct images in different planes and directions overcame the anatomical peculiarities of the maxillofacial region and made the positioning of the patient more adjustable during the examination.

7.2.5 Disadvantages of Computed Tomography

Computed tomography has several disadvantages. The presence of high-density structures, especially metallic structures such as metallic fillings, metallic frameworks of fixed prostheses, dental implants, etc., causes the hardening of the beam of X-rays and as a result the appearance of black zones in the image [15]. These black zones are called linear artefacts or artifacts and reduce the diagnostic ability of tomographic images by obscuring parts of the anatomical structures of interest or by simulating radiolucent pathological lesions. The radiation dose received by the patient is increased compared to the imaging modalities mentioned above, although low-dose imaging protocols have been developed in recent years without significant reduction in diagnostic information [16]. Finally, the increased cost of the examination is a disadvantage compared to the previous imaging modalities.

7.3 Cone Beam Computed Tomography

7.3.1 Principles of Operation

Cone beam computed tomography (CBCT) is an evolution of computed tomography. It initially found applications in angiography, radiotherapy, and mammography, and its use was subsequently extended to dental science [17–19]. The first cone beam scanner for use in dental science was installed in Europe in 1998 and later in the USA in 2001. It is a modern and now widely used imaging modality that uses a rotating tube that produces a cone-shaped beam of X-rays and is fixed opposite to the radiation detector system, thus showing several similarities with an orthopantomograph. The tube-detector system rotates typically 180–360° around the patient's head and takes successive tomographic images. The conical shape of the generated rays, the size of the cone, and the position and size of the detection system allows the tomograph to collect all the rays as they exit the various tissues of the patient and thus achieve scanning of the area of interest in a single rotation [20] (Fig. 7.1).

During its rotation the scanner takes 120–1024 two-dimensional images similar to lateral cephalometric radiographs as opposed to the axial sections found in computed tomography [21]. Total examination time, depending on the manufacturer and protocol used, ranges from 5 to 40 s.

Changing the shape of the beam of X-rays from a fan to a cone has increased the efficiency and reduced the mechanical complexity of the cone beam CT scanner compared to the conventional CT scanner, which scans the structures of interest step by step, producing axial sections of preselected thickness, therefore using more mechanical parts as well as increased examination time [22] (Fig. 7.2).

The production of a cone-beam computed tomography image is achieved in four stages: generating the beam and passing it through the structures of interest, collecting the radiation by the detection system, reconstructing the images, and viewing them [23].

7.3.2 The Beam Generation System

The beam is produced by a fixed-anode X-ray tube, which has similarities to the tubes used in orthopantomographs. The main difference

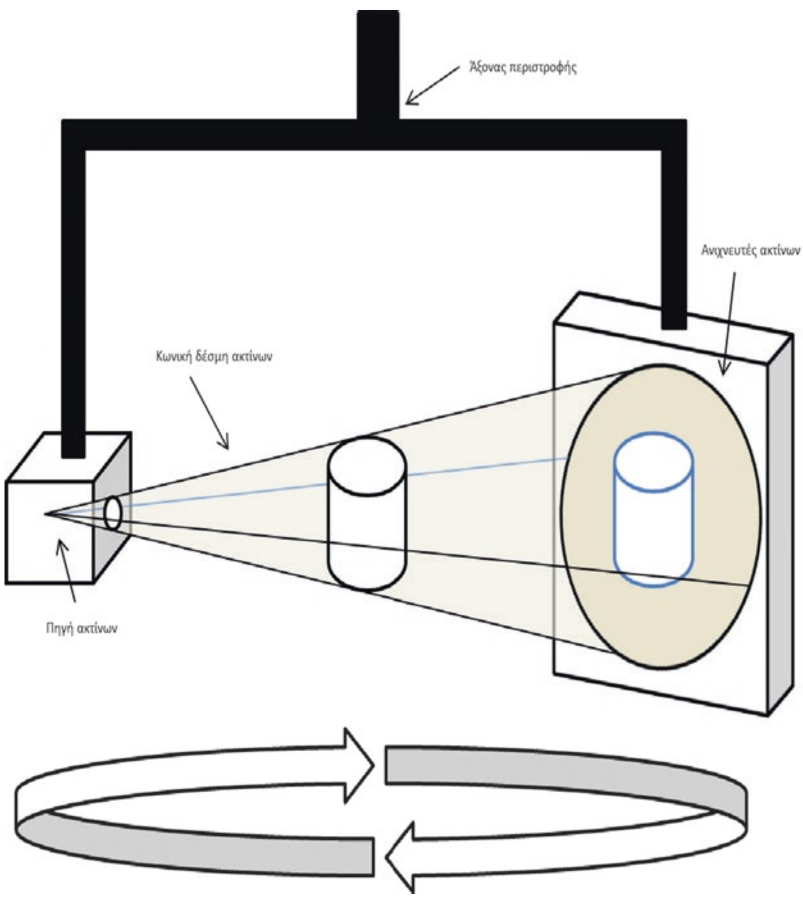

Fig. 7.1 The basic principle of rotation of the beam in CBCT machines

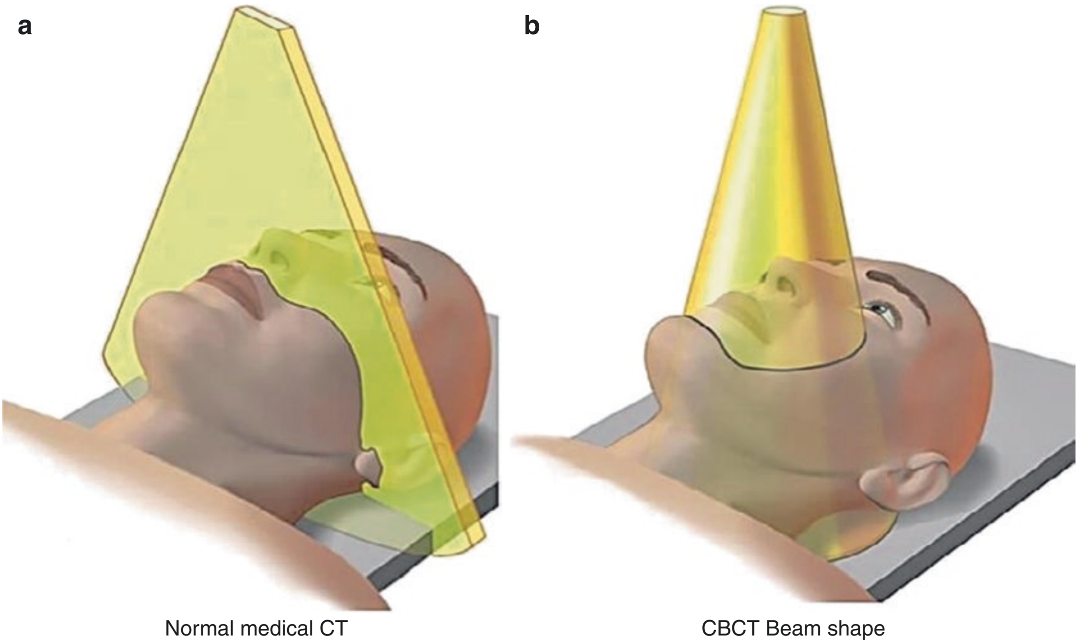

Fig. 7.2 The key difference of beam shape when compared to MDCT. (**a**) Normal medical CT and (**b**) CBCT beam shape

between the radiation system of the cone beam CT scanner and conventional CT is the introduction of beam confinement systems, which give the beam a cone shape with a base area approximately equal to that of the scanning system and limit the radiation exposure to the area of interest [24]. As the tube rotates around the patient, several images are produced, each corresponding to the projection of the structure of interest at the compared angle of irradiation. Due to the technical limitation of the detection system to receive the signal continuously, the beam emitted from the tube is not continuous but is interrupted (pulsed beam). The frequency of the pulsed beam is adjusted so that the interval between successive exposures coincides with the time required by the detection system of the scanner to process the data resulting from each exposure. Because of the pulsed beam, the patient's exposure time to radiation is much shorter than the total examination time [23, 24]. Each pulse produces a single projection image called the basal or structural image. As noted above, the basal images in cone-beam computed tomography are more similar to lateral cephalometric images than to basic axial sections generated by conventional CT. In most modern CT scanners the number of structural images produced is constant. A larger number of structural images in a cone-beam computed tomography scan results in an increase in the signal received by the detection system and consequently an improvement in the quality and sharpness of the final tomographic images due to an improvement in the signal-to-noise ratio (SNR). However, increasing the number of structural images, as long as the acquisition rate remains constant, leads to increased patient irradiation time.

7.3.3 The Radiation Detection System

The collection of the attenuated rays after they exit from the structures of interest is performed by the detection system of the cone-beam CT scanner. The detection system may consist of image intensifier tubes or the more modern and less bulky flat panel detectors [23].

Image intensifier tubes usually consist of an array of scintillation screens, two cesium iodide intensifier plates, corrective lenses, and a charge coupled device (CCD) sensor array [23–25]. In a detection system of this type, the attenuated beam enters the vacuum tube and is converted into light using an amplifying "input phosphor." The bright signal is then converted in the photocathode into an electron beam which is accelerated through an electric field and converted back into a light signal by the use of a second output phosphor. The intensity of the final light signal is adjusted by a series of lenses and an optical iris. The now corrected light signal is converted in the CCD sensor array into an electrical signal [25]. The intensity of the electrical signal generated by the CCD sensor is proportional to the intensity of the beam incident on the input intensifying plate. Tubular image intensifiers cause geometric distortion in tomographic reconstruction due to the shape of the vacuum tube and the many steps involved in acquiring the image, as well as increase the production of artefacts. The increase in noise in this type of scanning system is addressed by corrective mathematical algorithms in the scanner software [23–25]. Additional disadvantages of tubular image intensifiers include their increased volume, the need for more frequent calibration due to their sensitivity to magnetic fields and the fading of the intensification plates over time and use [26].

The flat plate detection system also uses an iodide cesium iodide amplifier plate to convert incident radiation into a light signal which is then captured and converted into an electrical signal by sensor arrays contained in a thin layer of amorphous silicon (a-si). The electrical signal produced by the sensors has an intensity proportional to that of the light signal, which in turn is proportional to the intensity of the radiation incident on the amplifying plate. These systems do not cause the geometric distortion and some of the artefacts found in tubular image intensifiers [5] and consequently reduce noise levels in tomographic images. The disadvantages of these systems are their non-linear sensitivity to low or high radiation intensities, the attenuation of the enhancement plate, and the uneven performance of their radiation detection capability exhibited at different points on their surface.

In both types of detectors, the dimensions and number of cells collecting the signal have a very large role in the analytical capability of the detection system. As the area of the sensitive cells is reduced, the analytical capability (resolution in space) of the detection system increases but at the same time the signal-to-noise ratio (SNR) decreases. The need for signal transport subsystems on the surface of the detectors limits the number of cells and therefore reduces the analytical capability of the system. In a comparison of the two types of detection systems, tomographic images produced with flat plate systems had higher resolution [24]. Finally, in contrast to computed tomography, the size of the voxel and therefore the resolution of the image in cone-beam computed tomography depends on the size of the sensitive cells of the detection system and not on the section thickness selected.

7.3.4 Reconstruction of the Tomographic Images

The process of converting the electrical signal generated by the detector system into an image is a complex process performed by the scanner's computer. First, a pre-correction of the inhomogeneity of the structural images produced is performed. This lack of homogeneity is the result of the different sensitivity levels of the cells included in various parts of the surface of the detection system and on of the inhomogeneity of the efficiency of the intensifying plate. After pre-correction, the computer, using complex mathematical algorithms, the most common being Feldkamp's algorithm or Feldkamp–Davis–Kress (FDK) method [27], sorts and corrects the data from the original structural images and finally displays them with a pixel brightness diametrically opposite to the original one. This technique is called a back projection technique.

7.3.5 Viewing the Scan Images

The cone-beam CT scanners project the tomographic images reconstructed from the original structural images in the axial, frontal, or ocular plane. These images are called secondary reconstructed images. The user can select the slice thickness of each reconstruction, enhance the resulting images with a number of tools available in the scanner software, as well as perform measurements and pre-operative planning. Particularly useful is the ability of cone-beam scanners to store and use all the scan data and to reconstruct images in other planes beyond the three basic ones depending on the user's diagnostic needs (multi-planar reformatting). Finally, as in conventional computed tomography, using specific algorithms, the scanner's computer creates shadows by illuminating the projected volumes in such a way as to give a three-dimensional appearance to the resulting image.

7.3.6 Advantages of Cone Beam Tomography

Cone beam computed tomography has a number of advantages compared to other imaging modalities. The scanning and data acquisition time is short (5–40 s), reducing the possibility of image distortion due to physical or respiratory movements of the patient. In modern cone-beam scanners the isotropic voxel size ranges from 80 to 400 μm and depicting the structures of the maxillofacial region with sufficient resolution, and thus allowing the operator to obtain measurements accurately. When compared to conventional computed tomography, cone beam computed tomography shows significantly fewer artifacts due to metallic structures in the region of interest [28] and the radiation dose to the patient (effective dose), depending always on the scanner, protocol and field size, is lower [28]. An important advantage is also the interactivity, i.e. the ability of modern systems to allow the transfer of tomographic data for study in other computer systems. Finally, the size and the cost of a cone beam computed tomography scanner is significantly reduced compared to conventional computed tomography scanners.

7.3.7 Limitations: Disadvantages of Cone Beam Tomography

In cone beam computed tomography a large part of the patient's head is irradiated due to the conical nature of the beam of X-rays produced. This fact statistically increases the probability of photon scattering and consequently the production of secondary radiation, which is recorded in the scanner's detection system as noise.

As a result of this phenomenon there is a reduction in the resolution and diagnostic capability of the resulting tomographic images. In conventional CT scanners, the fan-shaped beam improves the secondary (scatter radiation) to primary radiation ratio (SPR), compared to cone beam scanners. The SPR fraction has a value of 0.01 for single beam computed tomographs, 0.05–0.15 for fan-beam computed tomographs with helical scanning, and 0.4–2 for cone beam computed tomographs [23]. Finally, increasing the angular range of the cone beam leads to a larger area of the cone base, a larger imaging field, and consequently increased secondary radiation and noise generation.

As mentioned before, the production of artifacts is reduced compared to conventional computed tomography, yet their presence continues to affect the diagnostic ability of tomographic reconstructions despite the efforts of manufacturers to reduce them using improved mathematical algorithms. Artifacts are produced by differences between the mathematical model used by the scanner for the 3D image reconstruction and the physical properties of the various elements (detection system, object to be examined) involved in it. In cone beam computed tomography, the relevant artifacts are those caused by beam hardening, the partial volume averaging artifacts, the patient motion artifacts, the extinction artifacts, and the ring and aliasing artifacts.

7.3.8 Applications of Cone Beam Tomography

Cone beam computed tomography has a multitude of uses in dental science. In the surgery of the maxillofacial region it provides information for diagnosis and surgical planning needed to treat fractures and deformities, surgical extraction of impacted teeth and their relationship to structures of interest (maxillary canal, sinus, roots of adjacent teeth) and identification and determination of the location of foreign bodies [24–27]. In implantology, cone-beam computed tomography contributes to 3D preoperative planning and assessment of implant sites, assessment of peri-implant tissue health and planning of possible corrective procedures after the prosthetic rehabilitation stage, as well as the construction of surgical guides for guided implant placement.

In recent years, the usefulness of cone beam computed tomography in orthodontics as well as in endodontics has been studied.

7.3.9 Guidelines in Cone Beam Computed Tomography

Cone beam computed tomography, like any form of imaging examination using ionizing radiation, should follow a set of guidelines aimed at protecting the patient. A recent study speculates that in the USA up to 2% of recorded cases of malignancy may be due to radiation exposure from CT scans. The increase in the number of cone-beam CT scanners and their uses in more and more fields in dental science has led to efforts, both in the United States and Europe, to establish guidelines for indications for use, protocols and the conduct of examinations with cone-beam technology.

Initially, the American Academy of Oral and Maxillofacial Radiology in 2008, based on the ALARA (as low as reasonably achievable) principle, proposed in a related article a series of recommendations on how to use cone beam computed tomography, the responsibilities of the examiner, the basic principles of radiation protection and equipment quality assurance, and finally the justification of each scan. Similarly, in Europe, the SEDENTEXCT project (Safety and Efficacy of a new and emerging Dental X-ray modality) was developed with the contribution of the European Atomic Energy Commission (EURATOM) and the European Association of Dental Radiology (EADMFR). The project has published a set of guidelines including the basic principles of using cone beam computed tomography, suggested modalities for its application in the maxillofacial region, radiation dosage, and radiation protection for patients and medical staff.

SEDENTEXCT expired in 2011 and in May 2012 the guidelines published in a related article (Radiation Protection: Cone beam CT for Dental and Maxillofacial Radiology. Evidence based guidelines) were incorporated with the European Union radiation protection series.

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