P Sukovic

Cone beam computed tomography in craniofacial imaging

Author's affiliation:
P. Sukovic, Xoran Technologies, Ann Arbor, MI, USA

Correspondence to:
P. Sukovic
Xoran Technologies, Inc.
309 N. First St.
Ann Arbor, MI 48103
USA
Tel.: +1 734 663 7194
Fax: +1 734 663 8500
E-mail: psukovic@xorantech.com

Abstract

Author **–** Sukovic P

Cone beam computed tomography (CT) has the potential to reduce the size and cost of CT scanners. Because this emerging technology produces images with isotropic submillimeter spatial resolution, it is ideally suited for dedicated dentomaxillofacial CT scanning. When combined with application-specific software tools, cone beam computed tomography can provide dentomaxillofacial practitioners with a complete solution for performing specific diagnostic and surgical tasks, such as dental implant planning. In this paper, we provide a brief overview of cone beam scanning technology and compare it with the fan beam scanning used in conventional CT scanners. We introduce 'DentoCAT™', a relatively small, low-cost cone beam CT scanner dedicated for dentomaxillofacial imaging developed at Xoran Technologies. We present images generated by the DentoCAT™ scanner and provide an assessment of its performance in terms of spatial resolution and effective radiation dose. Finally, we illustrate the clinical utility of the scanner by presenting the results we have obtained to date using the DentoCAT™ scanner in conjunction with an implant planning software tool.

Key words: cone beam; computed tomography; CBCT

Introduction

Various imaging modalities have been used in the dentomaxillofacial fields over the past few decades, none of them with entirely satisfactory results. This is particularly true for more demanding imaging tasks, such as implant planning, temporomandibular joint

To cite this article:
Orthod. Craniofacial Res. **6** (Suppl. 1), 2003; 31–36
P. Sukovic:
Cone beam computed tomography in craniofacial imaging

ISSN 1741-2420

imaging, detection of facial fractures, lesions and diseases of soft tissue in the head and neck, and reconstructive facial surgery.

In particular, the use of dental implants is becoming the treatment of choice for the replacement of missing teeth. The successful outcome of a dental implant – the osseointegration of the implant – is heavily dependent on precise pre-surgical planning. Since the functional load on implants can be high, it is important that the implant be placed in a position where it can contact cortical bone and at an angle where the forces are as perpendicular as possible. Selection of the appropriate size and inclination of the implant in both a bucco-lingual and mesiodistal direction requires precise knowledge of the anatomy of the proposed site, including its dimension in all planes, the presence of knife-edge ridges and undercuts, as well as the location of anatomic structures, such as the nasal fossae, the maxillary sinus, and the mandibular canal (1). An evaluation of the thickness of the cortical bone and the density of the medullary bone is also critical to the success of the implant.

Commonly used dentomaxillofacial imaging modalities, such as periapical radiography, panoramic radiography, and conventional tomography produce only two-dimensional and/or distorted images. As a result, a number of practitioners have resorted to outsourcing computed tomography (CT) scans for implant planning and other demanding imaging tasks.

Principles of X-ray computed tomography

The CT scanners consist of an X-ray source and detector mounted on a rotating gantry (Fig. 1). During

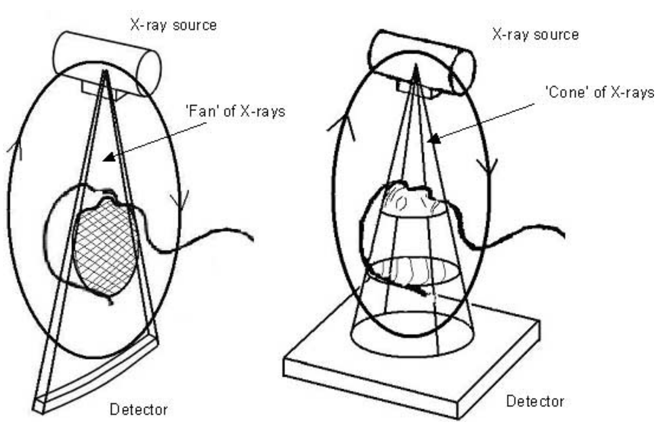

Fig. 1. Principles of conventional (fan beam) CT (left) and cone beam CT (right).

one rotation of the gantry, the detector detects the flux, I, of X-rays that have passed through the patient. The attenuation of monochromatic X-rays in homogenous objects is governed by: $I = I_0 \exp(-\mu x)$, where I_0 is the X-ray intensity without the object, x is the length of the X-ray path through the object, and μ is the linear attenuation coefficient of the material at the X-ray energy employed. For inhomogeneous objects, like the human body, the attenuation of X-rays consequently can be described by: $I = I_0 \exp(-\int\mu(x)\mathrm{d}x)$. By taking the logarithm of the flux, $-\log(I/I_0)$, one obtains line integrals of the linear attenuation co-efficients. These integrals constitute so-called 'raw data' that are then fed into an image reconstruction method that generates cross-sectional images whose pixel values correspond to linear attenuation coefficients. The theoretical background for tomographic image reconstruction was laid out in 1917 when Radon established that a three-dimensional object can be reconstructed from an infinite set of two-dimensional projections obtained at varying angles around the object.

The resulting attenuation coefficients are usually expressed with reference to water, and are given in Hounsfield units (HU):

$$\mathrm{HU}_{\mathrm{patient}} = 1000 \times \frac{\mu_{\mathrm{patient}} - \mu_{\mathrm{water}}}{\mu_{\mathrm{water}}}$$

The first CT scanner was developed in 1967 by Sir Godfrey N. Hounsfield, an engineer at EMI. Since then, CT technology rapidly underwent four developmental generations. The first generation of CT scanners used a single detector element to capture a beam of X-rays, corresponding to the integral of linear attenuation coefficients along a single line. It then translated horizontally to acquire the next line integral. After acquiring all the line integrals for a given position of the X-ray source, both the detector and source rotated one degree – a design known as the 'translate-rotate' or 'pencil-beam' scanner. Hounsfield's unit belonged to this generation, as did the first commercial CT scanners introduced in 1972. Interestingly, these first generation CT scanners were designed to scan the head only.

A second generation of CT systems was introduced in 1975. These systems, also known as 'hybrid' machines, used more than one detector and used small fan-beam, as opposed to pencil-beam, scanning. Like the first generation of CT scanners, these scanners also used a translate-rotate design, and most were head only scanners. While the first iterations of full body CT scanners also incorporated the translate-rotate design, image quality was poor because of patient motion artifacts caused by the significant amount of time required to take the scan.

Third generation CT scanners appeared in 1976 and are the systems most widely used today. These scanners use a large, arc-shaped detector that acquires an entire projection without the need for translation. This rotate-only design, frequently referred to as 'fan-beam', utilizes the power of the X-ray tube much more efficiently than the previous generations.

Fourth generation scanners shortly followed third generation scanners, replacing the arc-shaped detector with an entire circle of detectors. In this design, the X-ray tube rotates around the patient, while the detector stays stationary. Since these fourth generation scanners tend to be more expensive and suffer from higher levels of scatter, most of the commercially available CT scanners today are third generation scanners.

After an initial period of rapid development, CT technology quickly became mature, and it was not until the early 1990s that CT research began anew. Recent advances in CT include multirow detectors and spiral scanning. Multirow scanning allows for the acquisition of several cross-sectional slices at the same time, reducing scanning times. Today's state-of-the-art scanners have 16 rows of detectors. Spiral (helical) scanning incorporates a moving table with the rotating X-ray tube, with the net effect that the X-ray tube describes a helical path around the patient.

Conventional CT scanners are large and expensive systems designed primarily for full-body scanning at a high speed to minimize artifacts caused by movement of the heart, lungs, and bowels. They are not well-suited for in house use in dentomaxillofacial facilities, where cost considerations are important, space is often at a premium, and scanning requirements are limited to the head. The advent of cone beam computed tomography (CBCT) technology has paved the way for the development of relatively small and inexpensive CT scanners dedicated for use in dentomaxillofacial imaging.

Principles of cone beam computed tomography

The CBCT scanners utilize a two-dimensional, or panel, detector (Fig. 2), which allows for a single rotation of the gantry to generate a scan of the entire head, as compared with conventional CT scanners whose multiple 'slices' must be stacked to obtain a complete image. Cone beam technology utilizes X-rays much more efficiently, requires far less electrical energy, and allows for the use of smaller and less expensive X-ray components than fan-beam technology. In addition, the fan-beam technology used in conventional CT scanners does not lend itself to miniaturization because it requires significant space to spiral around the entire body.

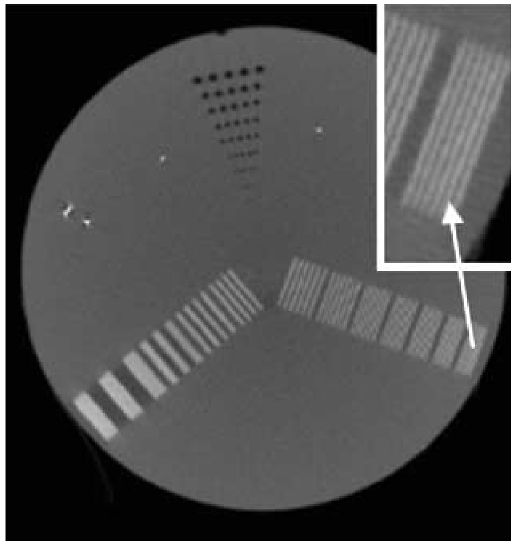

Fig. 2. Image of the bar pattern insert of the quality assurance phantom. The enlarged area of the bar pattern has 11 lp/cm, which is clearly visible.

Jaffray and Siewerdsen noted in (2): 'the CBCT approach offers two important features that dramatically reduce its cost in comparison to a conventional scanner. First, the cone beam nature of the acquisition does not require an additional mechanism to move the patient during the acquisition. Second, the use of a cone beam, as opposed to a fan beam, significantly increases the X-ray utilization, lowering the X-ray tube heat capacity required for volumetric scanning. For the same source and detector geometry, the efficiency roughly scales with slice thickness. For example, the X-ray utilization increases by a factor of 30 in going from a 3 mm slice in a conventional scanner to a cone angle corresponding to a 100 mm slice with a cone beam system. This would reduce heat load capacity dramatically. From our experience, a 5200 KHU X-ray tube costs approximately $70,000, whereas a 600 KHU X-ray tube (a factor of ~10 lower in capacity) costs roughly $6000'.

Because the head and neck can be sufficiently stabilized for clear imaging at a slower scanning speed, a dedicated dentomaxillofacial scanner does not require the highly sophisticated, bulky, and expensive components required for sub-second scanning in full-body CT scanners to avoid blurring of the images caused by movement of the heart, lungs, and bowels.

In short, CBCT is ideally suited for high quality and affordable in-house or on-site CT scanning of the head and neck in dentomaxillofacial applications.

This value of using CBCT for dedicated dentomaxillofacial imaging has been recognized by a number of researchers, and several commercial systems are in development or are already available from Quantitative Radiology, Verona, Italy (3) (NewTom 9000), Hitachi and Morita Co., Tokyo, Japan (4, 5) as well as Xoran Technologies, Ann Arbor, MI, USA (6) (DentoCAT^{TM}). In the following section, we present some of the preliminary results obtained with the prototype of Xoran's DentoCAT^{TM} scanner.

Initial performance assessment

We used two methods to estimate spatial resolution of the DentoCAT^{TM} scanner. First, we visually assessed the images of a spatial resolution bar pattern insert of the Mark II Quality Assurance Phantom. Second, we estimated the modulation transfer function (MTF) of the system from a steel wire within the same insert. Figure 2 shows a reconstructed image of the spatial resolution bar pattern insert. The image was reconstructed on a grid of 0.2 mm pixel size with Hanning filter with a cut off frequency of 0.9.

We calculated the MTF of the system to more accurately assess the spatial resolution properties of our scanner. The MTF was calculated as the absolute value of the normalized complex Fourier transform of the line spread function (LSF). The LSF has been measured directly by scanning a steel wire of the diameter of 0.1 mm. The wire was positioned approximately 5 cm from the axis of rotation.

To minimize the effects of pixel size on MTF, the image was reconstructed on a very fine grid of 0.05 mm pixel size. As the wire is not infinitely thin, the 'raw' MTF must be corrected for the finite size of the wire. We accomplished this correction by dividing the 'raw' MTF by the spatial frequency distribution for the wire. From the plot of MTF the following values were assessed for the central slice: MTF (2%) = 15 lp/cm, MTF (10%) = 12 lp/cm, and MTF (50%) = 6.5 lp/cm. The values for the off-central slice were only slightly degraded.

In assessing the effective dose, we closely followed the methodology suggested in Publication 60 of the International Commission on Radiological Protection (ICRP) (7), and Frederiksen et al. (8). Twenty thermoluminescent detectors (TLDs) were placed in selected sites representing radiosensitive tissues or organs in the Rando Head phantom. The X-ray tube settings were the same as in the experiments described above – tube potential was set to 110 kVp, the beam was filtered with 0.4 mm of Cu and 1 mm of Al. The total effective dose was calculated to be 0.585 mSv, which is below the values for effective dose typically obtained with conventional CT maxilla and mandible protocols.

Clinical utility

The need for accurate imaging for implant planning is a compelling example. Recently, novel CAD–CAM techniques, such as stereolythographic rapid prototyping, have been developed to build surgical guides based on CT scans for the purpose of improving the precision of implant placement. Below we illustrate the utility of using DentoCAT^{TM} with one such system (Materialise Inc., Leuven, Belgium), and compare it to a conventional surgical guide. The results of this work were originally reported in (9).

The experiment is outlined in Fig. 3. Five epoxy mandibular edentulous jaws were scanned and implant planning was performed using a commercially available software package, Surgicase (Materialise Inc.). Five surgeons performed osteotomies: on the right sides of the jaws, they utilized a conventional surgical guide and on the left sides, they used a stereolythographic guide that was custom designed based on the CT scan and implant planning procedure. The jaws were then scanned again and an image registration software package (Analyze 4.0, Lenexa, KS, USA) was used to register the pre- and post-operative scans. This allowed for the display of the planned (virtual) implants and osteotomies in the same image as well as for

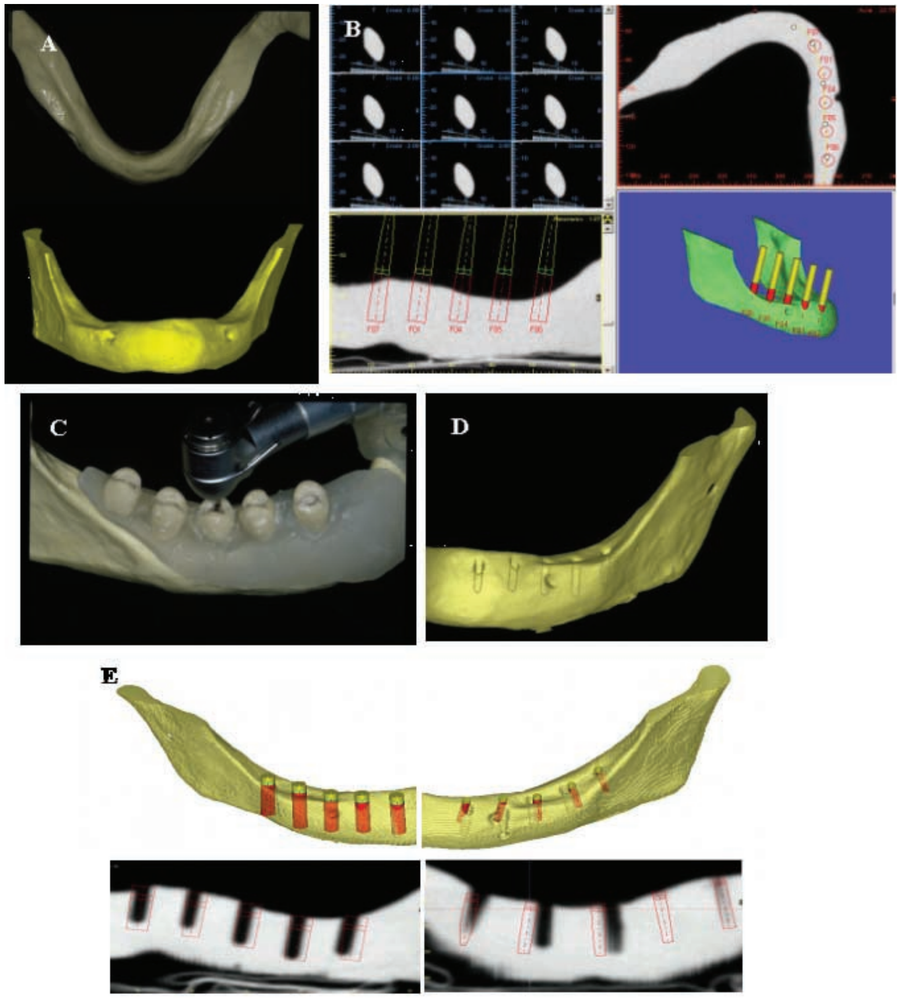

Fig. 3. (A) Photograph and three-dimensional rendered CT scan of an epoxy jaw. (B) Implant planning is performed by placing virtual implants on the right side where a standard surgical guide is to be used, and on the left side where the test guide will be used. (C) Based on the planning, surgical guides are fabricated and osteotomies are performed. (D) 'Post-surgical' scan of the jaws are performed. (E) The pre- and post-surgical scans of the jaws are co-registered and the errors measured and compared with the control study on the right.

measurements of inaccuracies. The average distance between the planned implant and the osteotomy was 1.5 mm at the entrance and 2.1 mm at the apex using conventional surgical guides. The corresponding errors were significantly reduced to 0.9 and 1 mm using the stereolythographic templates.

In summary, cone beam CT is a versatile emerging technology whose high and isotropic spatial resolution, undistorted images, compact size and relatively low cost, make it a perfect candidate for a dedicated dentomaxillofacial imaging modality. When combined with dedicated software packages, it can provide practitioners with a complete solution for demanding tasks, such as implant planning.

Acknowledgements: This work is funded in part by NIH R43 DE14104-01, NIH R01 CA65637, and the University of Michigan. The author is indebted to Drs Sharon Brooks and David Sarment for their valuable input on this project.

References

1. Tyndall DA, Brooks SL. Selection criteria for dental implant site imaging: a position paper of the American Academy of Oral and Maxillofacial Radiology. *Oral Surg Oral Med Oral Pathol Oral Radiol Endod* 2000;**89**:630–7.
2. Jaffray DA, Siewerdsen JH. Cone-beam computed tomography with a flat-panel imager: initial performance characterization. *Med Phys* 2000;**27**:1311–23.
3. Mozzo P, Procacci C, Tacconi A, Martini PT, Andreis IA. A new volumetric CT machine for dental imaging based on the cone-beam technique: preliminary results. *Eur Radiol* 1998;**8**:1558–64.
4. Arai Y, Tammisalo E, Iwai K, Hashimoto K, Shinoda K. Development of ortho cubic super high resolution CT (Ortho-CT). In: Lemke HU, Vannier MW, Inamura K, Farman A, editors. *Computer Assisted Radiology and Surgery*. Amsterdam: Elsevier; 1998. pp. 780–5.
5. Arai Y, Tammisalo E, Iwai K, Hashimoto K, Shinoda K. Development of a compact computed tomographic apparatus for dental use. *Dento-Maxillo-Facial Radiology* 1999;**28**:245–8.
6. Sukovic P, Brooks S, Perez L, Clinthorne NH. *DentoCAT – A Novel Design of a Cone Beam CT Scanner for Dentomaxillofacial Imaging: Introduction and Preliminary Results*. 15th International Congress

and Exhibition on Computer Assisted Radiology and Surgery, 27–30 June 2001, Berlin, Germany; 2001.

7. International Commission on Radiological Protection. *Radiation Protection. 1990 Recommendations of the International Commission on Radiological Protection*. Oxford: Pergamon Press; 1990.
8. Frederiksen NL, Benson BW, Sokolowski TW. Effective dose and risk assessment from computed tomography of the maxillofacial complex. *Dento-Maxillo-Facial Radiol* 1995;**24**:55–8.
9. Sarment DP, Sukovic P, Clinthorne N. Accuracy of implant placement with a stereolythographic surgical guide. *J Oral Maxillofac Implants* (in press).