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Tilting the jaw to improve the image quality or to reduce the dose in cone-beam computed tomography

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ABSTRACT

Objective: The image quality in cone-beam computed tomography (CBCT) should be improved tilting the mandible that contains two dental titanium implants, within the relevant range of motion.
Materials and methods: Using the mandible of a five-month-old pig, CBCT was performed varying the accelerating voltage, beam current, the starting rotation angle of the mandible in the source-detector plane and the tilt angles of the jaw with respect to the source-detector plane. The different datasets were automatically registered with respect to micro CT data to extract the common volume and the deviance to the pre-defined standard that characterizes the image quality.
Results: The variations of the accelerating voltage, beam current and the rotation within the source-detection plane provided the expected quantitative behavior indicating the appropriate choice of the imaging quality factor. The tilting of the porcine mandible by about 14° improves the image quality by almost a factor of two.
Conclusions: The tilting of the mandible with two dental implants can be used to significantly reduce the artifacts of the strongly X-ray absorbing materials in the CBCT images. The comparison of 14° jaw tilting with respect to the currently recommended arrangement in plane with the teeth demonstrates that the applied exposure time and the related dose can be reduced by a factor of four without decreasing the image quality.

1. Introduction

Dentomaxillofacial X-ray cone-beam computed tomography (CBCT) has been applied to visualize the hard tissues for planning the implant placement and for the selection of the implant [1–4]. Based on these data, the oral surgeon can precisely identify the location of the mandibular nerve and other important structures [1] and decide subsequently, which part of the bony tissue will be removed for the implant placement and which part of the bone is needed to guarantee the mechanical stability of the patient's mandible. Unfortunately, CBCT cannot be utilized likewise for the post-operative imaging as validation and follow-up, because massive artifacts that appear as dark bands and streaks cover the region around the metal implant [5] that are also well-known from other metallic implants [6,7]. As consequence, the post-operative imaging only relies on radiographs. Because of the high X-ray absorption of the implant material, the number of transmitted photons available to visualize the surrounding tissue is extremely low, so that the artifact formation becomes easily understandable. The X-ray absorption also explains the orientation of the artifacts from the implant towards the stronger absorbing teeth and bony components. This behavior is especially evident between distinct, strongly absorbing implants.

The present communication deals with the possibilities of the CBCT operator to reduce the artifacts caused by standard dental titanium implants. First, the operator can increase the accelerating voltage until the maximum value has been reached. The increase is supportive, since the implants get more and more transparent for the photons of higher and higher energy. Admittedly, the soft and hard tissues of interest also become more and more transparent. Therefore, the improvement of the overall image quality is expected to be only limited or even absent. Second, the operator can prolong the exposure time or increase the beam current, which leads to a linear increase of the number of available photons but which also linearly raises the dose for the patient. The benefit for the signal-to-noise ratio only improves by the square root of the exposure time. Hence, the prolongation of the exposure time will not really improve the entire imaging procedure. Third, we have realized that using CBCT the head of the patient is oriented that teeth, potential

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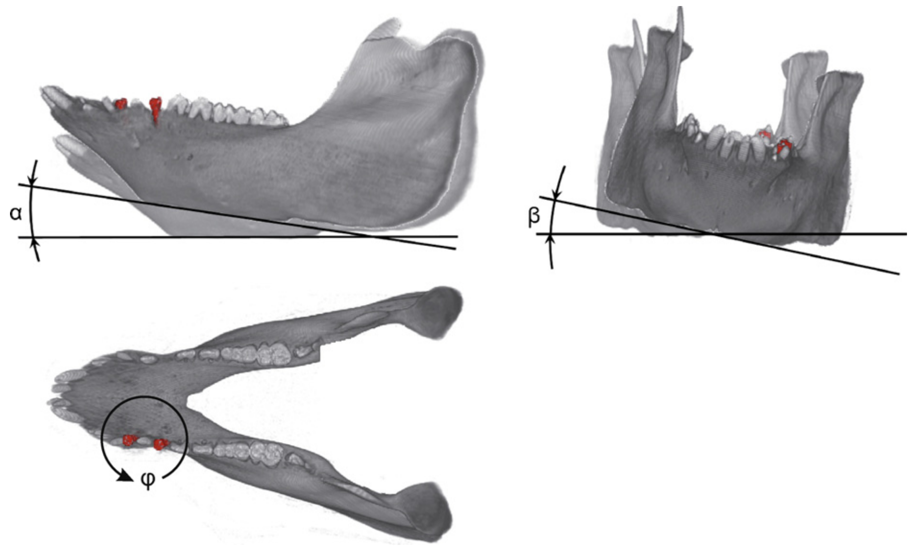

Fig. 1. The red color indicates the position of the two titanium implants in the 3D representations of the porcine mandible. The angles α and β denote the tilt of the mandible with respect to the source-detector plane of the CBCT in frontal and sagittal planes, respectively.

implants as well as X-ray source and detection unit are arranged in a single plane. This choice, however, is the worst case, since the highly X-ray absorbing components in line cause the strongest possible artifacts. Therefore, we hypothesize that the tilting of the head significantly reduces these artifacts. The present study should clarify how far the tilting of the head within the reasonable range improves CBCT image quality. The effect should become especially clear for patients with multiple implants, because the tilting can prevent the overlapping of the highly X-ray absorbing materials in the projections to be recorded. Therefore, the improvement of the CBCT image quality by tilting the jaw containing two dental titanium implants should be quantified within a relevant range of motion. Micro computed tomography with conventional or synchrotron radiation sources performed on the cylindrical part of the porcine mandible containing the dental implants provides the almost artifact-free standard of comparison.

2. Materials and methods

2.1. Specimen preparation

To avoid any radiation damage of test persons, the lower jaw of a five-month-old, meliorated country pig has been selected for the post-mortem experiment. This choice should be an appropriate model to prove the hypothesis. The anatomy of the pig's jaw cannot be simply translated to humans, although mini-pigs are regarded as adequate model [8]. The tongue and pieces of the vertebral body were added [9] to built a more realistic situation. For the better handling, the entire specimen was laminated in plastics. Because the pig was in the late phase of deciduous dentition two gaps for implantation were found mesial and distal of the canine. Two Ti-implants (SLA-surface, Straumann AG, Villeret, Switzerland) each 4.1 mm in diameter and 10 mm long were inserted according to the recommendations of the manufacturer.

2.2. CBCT imaging

The 3D Accuitomo 60 (Accuitomo, Morita, Japan) served for the CBCT image acquisition. For all CBCT scans the exposure time corresponded to 17 s. The accelerating voltage was varied between 70 and 80 kVp and the X-ray beam currents between 1 and 10 mA to check their influence on the image quality. In order to determine the influence of the specimen position with respect to the source and detection unit, the specimen was manually rotated to $\varphi = 0^{\circ}$, $\varphi = 30^{\circ}$, $\varphi = 45^{\circ}$, and $\varphi = 90^{\circ}$ around the implant's axis and was tilted in the frontal and sagittal planes by the angles α and β as represented in Fig. 1.

The projection data were exported and reconstructed by means of the modified Feldkamp algorithm using the Skyscan Nrecon^{TM} software (Skyscan, Kontich, Belgium). The reconstructed data volume corresponds to a cylinder 6 cm in diameter and 6 cm high, which contains the two Ti-implants inserted.

2.3. Micro computed tomography

For the micro computed tomography (μCT) measurements, a cylinder of the porcine mandible, which included the two implants, was extracted by means of a 3 cm hollow drill.

First, this cylinder was scanned using a conventional system, i.e. SkyScan 1172^{TM} (SkyScan, Kontich, Belgium) using an accelerating voltage of 100 kVp (with Al/Cu filter), a beam current of 100 μA, an exposure time of 2.7 s per projection, and a pixel size of 17.4 μm in rotation steps of 0.4° between 0° and 360°. The reconstruction was carried out with the Skyscan software package.

Second, synchrotron radiation-based μCT (SRμCT) measurements were performed at the beamline W 2 (HASYLAB at DESY, Hamburg, Germany) by the standard setup in absorption contrast mode, operated by the GKSS Research Center [10]. Here, the X-ray beam is monochromatic. The photon energy was chosen to 76 keV and the pixel size to 6.7 μm. This parameter choice resulted in a spatial resolution of 14.6 μm, experimentally derived from the projection of a sharp edge of a highly X-ray absorbing material. It corresponds to the 10% value of the modulation transfer function [11].

Since the cylindrical specimen was almost 3 times larger than the field of view (1536 × 730 pixels and 10.4 mm × 4.9 mm, respectively) 1440 projections were acquired at asymmetric rotation axes and combined with pixel precision before reconstruction at two height levels [12]. Because of the limited number of projections acquired and to increase the contrast (density resolution) the data were binned with a factor of 2 before reconstruction [13]. For easier data handling, the data were also binned by a factor of two after reconstruction. Hence, the volume, reconstructed by means of the filtered back-projection algorithm, consists of 32.3 mm × 32.3 mm × 8.9 mm of isotropic 27 μm-wide voxels.

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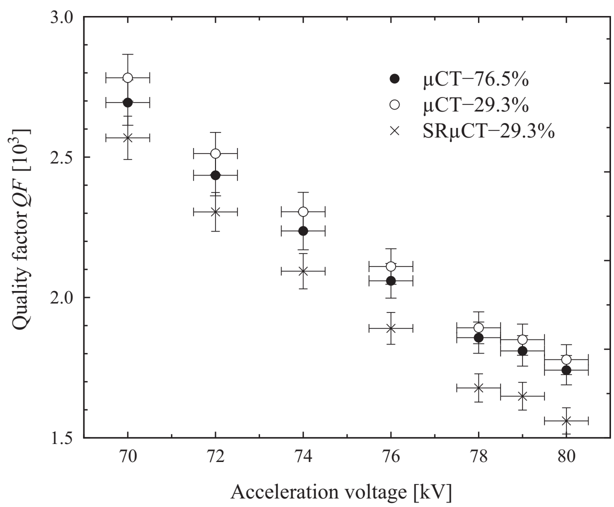

Fig. 2. The 3D image quality factor steadily decreases for acceleration voltages between 70 and 80 kVp. The behavior closely resembles for the CBCT data registered against μCT (common volumes 76.5% and 29.3% of the μCT dataset) and SRμCT, respectively.

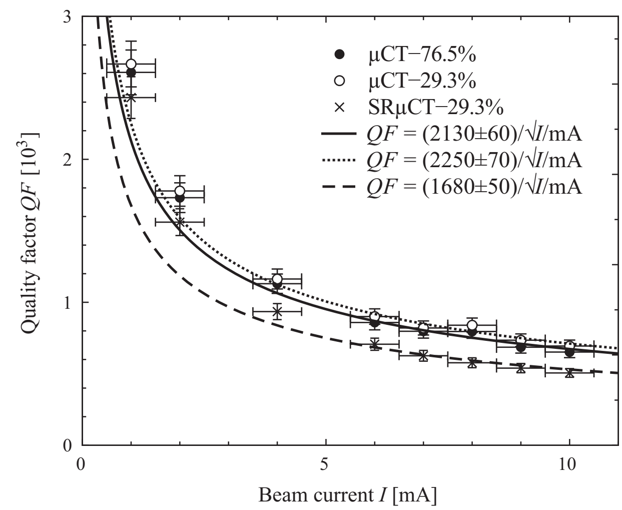

Fig. 3. The 3D image quality factor improves increasing the beam current according to the square root dependency. The CBCT data registered against μCT (common volume 76.5% and 29.3% of the μCT dataset) correspond to the ones registered against SRμCT.

2.4. Data processing

The software VG StudioMax 1.2.1 (Volume Graphics GmbH, Heidelberg, Germany) served for the tomography data visualization and the conversion of the CBCT projections from DICOM to RAW format.

For the quantitative analysis, the CBCT data were manually pre-registered with both μCT datasets as described previously [9,14]. The manual pre-registration procedure yields the starting values for the automatic rigid (affine) registration [15] subsequently performed. Similar registration procedures have been successfully applied previously [16].

Based on IDL 6.4 (ITT Visual Information Solutions, Bolder CO, USA), code was developed to crop the common volume for the direct comparison of the tomography data. Finally, the deviance of the detected local absorption values (gray levels) between the CBCT-data and the (SR)μCT-data used as standard for comparison were added up to quantify the image quality. The term 'image quality factor' – *QF* has been selected, since image quality refers to a characteristic value that measures the difference to an almost artifact-free standard. Hence, close similarity yields a low value for the quality factor, meaning higher image quality and less artifacts, respectively.

3. Results

It has been found that the internal reconstruction software of the 3D Accuitomo 60 system includes rather arbitrary background substitution and, therefore, leads to quality factors not reliably comparable, see [9]. Note, that the use of the internal reconstruction software gives rise to massive non-predictable quality factor modulations altering the starting angle φ. Consequently, the projections of the CBCT system were extracted and reconstructed using the Skyscan software package using one set of pre-defined parameters. Figs. 2 and 3 show the resulted 3D image quality factors (*QF*) as the function of the acceleration voltage (kVp) and of the beam current (mA) with the constant exposure time, respectively. Increasing the accelerating voltage towards the maximum possible value of 80 kVp the image quality improves continuously. This functional dependence arises for both the registration of CBCT against μCT and the registration of CBCT against SRμCT. Furthermore there is no strong habit on the size of the common volumes chosen, as demonstrated in Figs. 2 and 3 applying about 3/4 and 1/4 of the μCT volume. Image quality improvement is also found increasing the beam current. Here, the quality factor follows the square root of the beam current, see fits represented by the full, dashed, and dotted lines in Fig. 3.

The error bars for the 3D image quality factor were determined rotating the porcine jaw stepwise within the source-detector plane. The results for selected conditions are summarized in Table 1. They show that even the stepwise manual rotation of the mandible yields to 3D image quality factors that exhibit minor scattering around the mean values, namely maximal 6%.

Contrary to the rotation within the source-detector plane, tilting the pig's mandible with the two dental Ti-implants in the frontal

Table 1
Quality factors *QF* acquired at an acceleration voltage of 80 kVp and at the beam currents *I* for stepwise rotation along the CBCT axis φ.

I (mA)	φ (°)	*QF* (μCT – 76.5%)	*QF* (μCT – 29.3%)	*QF* (SRμCT – 29.3%)
2	0	1741.18	1778.64	1560.47
2	30	1701.43	1752.12	1521.10
2	45	1720.02	1750.12	1517.32
2	90	1763.69	1835.73	1608.37
7	0	785.61	820.42	626.04
7	30	780.18	824.46	626.75
7	45	833.95	852.62	615.17
7	90	789.57	828.34	631.90
10	0	643.51	694.19	505.87
10	30	657.17	709.27	537.40
10	45	666.33	712.62	520.52
10	90	655.31	708.12	529.78

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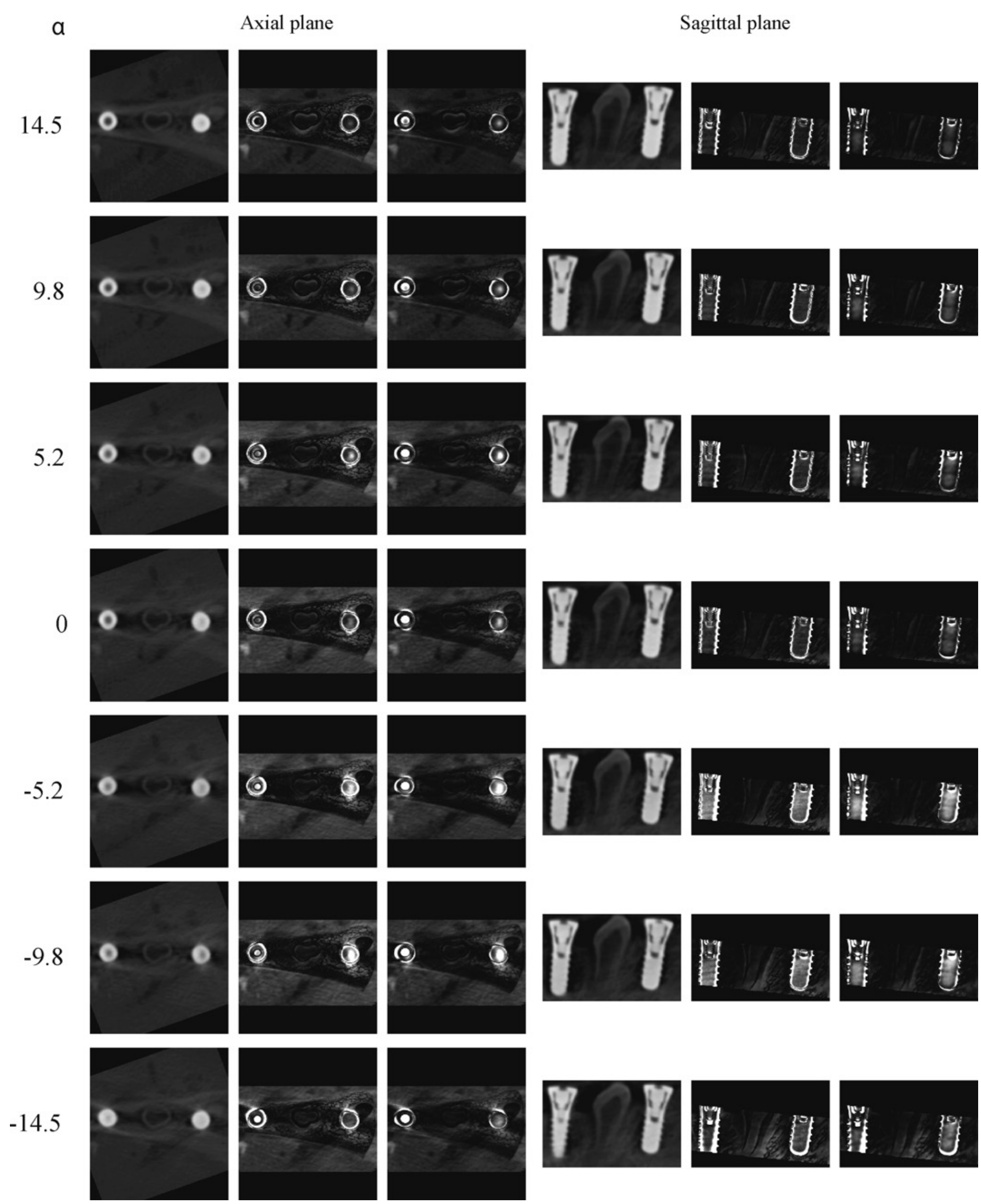

Fig. 4. The first and fourth columns are virtual cuts of the CBCT data in axial and sagittal planes, respectively. The images acquired with identical X-ray beam (80 kVp, 7 mA for 17 s) are labeled row-wise along with the tilt angle α. The second/fifth and the third/sixth columns represent the difference of the CBCT data and the registered common volumes with μCT and SRμCT. Less features means closer resemblance to the (SR)μCT and, therefore, better image quality (smaller *QF*) of CBCT.

and sagittal planes denoted as angles α and β leads to significant differences as qualitatively demonstrated by the image series in Fig. 4 and quantitatively summarized in Table 2. The results of the porcine mandible with the two implants incorporated indicate a stronger dependence for the angle α compared to the angle β.

4. Discussion

Although streak artifacts are well known in conventional and cone-beam CT, especially in tomography data of the orofacial region of patients with metallic dental fillings and implants, their quantitative assessment for the detailed improvement of the tomography data is lacking. The dependences of the quality factor versus accelerating voltage and beam current exhibit the predictable behavior and, therefore, demonstrate that the choice of the dedicated 3D image quality factor *QF* is an adequate measure of present streak artifacts. The selected quality factor is not primarily related to the appearance of the artifacts but rather a characteristic parameter for the similarity between the registered and subtracted 3D datasets of matching volume. Because the μCT data contain only imperceptible artifacts, similarity and quality are here well comparable quantities. Gross et al. [17] used a similar quantity to compare artifacts of differently composed clips namely the standard deviations of replicate scans. Although the measured standard deviations directly relate to the X-ray density of the materials, the quantitative description of the phenomena behind remains missing.

Tilting the head by certainly moderate angles can improve the 3D data quality by a factor of two, which is a substantial gain. Note that the quality improvement by a factor of two corresponds to the increase of beam current or dose to the patient by a factor of four. This means that tilting the jaw or introduc-

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Table 2
Quality factor *QF* for the tilt angles α and β accelerating voltage 80 kVp, beam current of 7 mA).

α (°)	β (°)	*QF* (μCT – 76.5%)	*QF* (μCT – 29.3%)	*QF* (SRμCT – 29.3%)
14.5	0	490.38	569.57	409.22
9.8	8.0	607.64	671.41	506.20
9.8	0	613.38	676.15	515.87
5.2	0	724.31	778.28	600.96
0	0	785.61	820.42	626.04
−9.8	0	816.43	868.46	681.42
−5.2	0	848.69	909.87	688.35
−14.5	0	843.93	934.60	694.45
0	14.5	872.59	947.37	718.72
0	8.0	898.55	954.61	714.84

ing another degree of freedom for the CBCT instrumentation such as angling the gantry avoiding the incidence of the highly X-ray absorbing implant(s) and teeth in the source-detector plane allows severely reducing the dose for the patient without loss of image quality.

The observation that the change of angle α improves the quality factor more than the variation of the angle β is associated with the special anatomy of the porcine mandible and the selected location and orientation of the two dental implants. Further research activities should fertilize the development of simple approaches that allow the operator of the CBCT system selecting the optimized tilting of the patient's head with respect to the source-detector plane.

The results can be directly converted to CT modalities. For the standard, however, μCT is necessary. Note CBCT systems provide comparable or even better image quality than multi-slice CT [18].

5. Conclusions

The deviance of absorption values in registered CT data is an appropriate parameter to characterize the similarity between selected standard and CT datasets introduced as 3D image quality factor. Both standards used, conventional μCT and SRμCT provide comparable results, which implies that the laboratory sources are powerful enough to serve as standard data sets. The error bar of the imaging quality factor can be extracted from stepwise rotation of the specimen within the source-detection plane. Moderate tilting of the porcine mandible containing two dental implants can improve the image quality by a factor of two.

Appropriate tilting the jaw containing highly X-ray absorbing implants can reduce the dose by a factor of four without the loss of image quality. Tilting the jaw to reduce the overlap of highly X-ray absorbing implants and hard tissues significantly reduces the related artifacts and thus contributes towards post-operative imaging for validation and follow-up.

Conflict of interest

No author has any conflict of interest.

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