RESEARCH ARTICLE

TLD measurements and Monte Carlo calculations of head and neck organ and effective doses for cone beam computed tomography using 3D Accuitomo 170

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Objectives: In dentistry, the use of cone beam CT has steadily increased over the last few years. The aim of this study was to measure organ doses and to perform dose calculations based on Monte Carlo (MC) simulations to work out a basis for full three-dimensional (3D) dose calculations for any patient examination performed with the machine used in this study.

Methods: TLD-100 LiF detectors were placed at 71 measurement positions on the surface and within a RT-Humanoid phantom to cover all relevant radiosensitive organs and tissues. Three examinations with different protocols were performed with the 3D Accuitomo[®] and dose calculations with MC simulations were carried out for the same three protocols using the EGSnrc MC transport code system.

Results: Field of views of 140×100 , 80×50 and 40×40 mm² were selected, the mean organ doses were measured as 5.2, 2.75 and 1.5 mGy and the effective doses were determined as 250, 97 and 48 µSv. For the MC simulation of organ doses and the thermoluminescent dosimeter measurements, an overall agreement within $\pm 10.1\%$ (two standard deviations) was achieved. The measured dose values for 3D Accuitomo[®] were about a factor 2 lower when compared with conventional CT examinations.

Conclusions: Reliable results for the organ doses as well as effective dose values were achieved with thermoluminescent dosimeter measurements in the RT-Humanoid phantom. This study provides the basis for the application of MC simulations for further dose determinations of cone beam CT machines. The MC calculation may therefore be a valuable tool to support the dentists in the evaluation of the trade-off between additional information that may be relevant to the choice of therapy and the additional dose given to the patient.

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Introduction

The aim of this study was to investigate the dosimetric properties of the cone beam CT (CBCT) device

3D Accuitomo 170 (J. Morita Mfg Corp. Kyoto, Japan) by dose measurements of organ doses and to develop and validate a dose calculation tool based on Monte Carlo (MC) methods. MC simulations are a well-established technique and the most accurate

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tool for patient imaging dose calculations.¹ With MC methods, the possibility arises to calculate patient-specific three-dimensional (3D) dose distributions.

As the need for 3D information acquisition in dentistry with CBCT is growing rapidly, the contribution to the collective dose through dental radiology will be increased. In general, diagnostic imaging is contributing to about 30% to the collective dose per year as demonstrated in a radiation protection survey in Switzerland in 2008.² The contribution of dental radiology examinations to the collective dose amounted to less than 1% of the total dose in comparison with CT applications with 68%. On the other hand, the contribution of dental radiology with respect to the frequency of examinations reached 42% of the total number of examinations per year and was almost as high as the contribution of medical radiography with 46%.² When the contribution to the collective dose through dental radiology will be increased, it is essential to be able to quantify the corresponding risks of this radiation exposure with appropriate accuracy as well to have a tool for patient dose calculation as well as optimization, which results in the motivation for this study. Regarding the evaluation of a patients radiation exposure and validation of dose reduction models,³ 3D dose distributions as obtained with MC calculations are more valuable than single-point measurements with regard to the evaluation of organ doses. Single-point measurements are only feasible at very few locations within a patient as well as they are limited in number of locations within an organ. MC calculation may therefore be a valuable tool to support the dentists in the evaluation of the trade-off between additional information that may be relevant to the choice of therapy and the additional dose given to the patient.

Since in this work an MC method was used for the estimation of dose distributions from a CBCT device, it was possible to investigate parameters like field size [field of view (FOV)] or imaging protocol with respect to dose optimization as well as patient dose estimation without the exposure of a patient.

Regarding the experimental validation of the developed MC model, many different approaches exist. To describe the dosimetry for CBCT, the device specifications such as volume computed tomography dose index, dose length product or dose area product may be assessed with ionization chamber measurements.⁴ Although these device-specific measurements could be taken as an MC validation, they do not provide any additional patient-related dosimetric information. Film measurements can be used as simple measurement device to get information about two-dimensional dose distributions. However, these are limited in the location as well as orientation of the measurement. In many publications about CBCT dosimetry,^{3,5–7} measurements in anthropomorphic phantoms with thermoluminescent dosimeters (TLDs) have been used due to their small size, simple handling and good reproducibility. In this

study, TLD dosimetry in an anthropomorphic phantom was chosen for the MC model validation to reproduce the conditions during exposure of a patient as realistic as possible and in addition to be consistent with a previous study of Dula *et al*⁸ conducted with the same anthropomorphic phantom.

Methods and materials

All measurements and calculations were done for the 3D Accuitomo 170 (J. MORITA Mfg Corp.).⁹ It consists of a D-051 Toshiba (Toshiba Electron Tubes & Devices Co., Otawara, Japan) X-ray tube (60–90 kV, 1–10 mA) and a flat panel detector with an FOV ranging from 40 × 40 to 170 × 120 mm². The specifications of 3D Accuitomo 170 with respect to dosimetry are summarized in Table 1.

Thermoluminescent dosimeter measurements

For all measurements, cubic TLD-100 chips (LiF: Mg,Ti) from Harshaw (Harshaw/Bicron Radiation Measurement Products, Solon, Ohio) TLD Bicron were used with dimensions of 3 × 3 × 0.9 mm³. The heating cycle consisted of an annealing phase (400°C for 1 h, 100°C for 1 h) with a passive cool-down process, a pre-heating phase (100°C for 15 min) to prepare the TLDs and a readout process (from 50 to 350°C with 15°C s⁻¹). The Heraeus AG (oven type K570) and the Harshaw reader 5500 (Thermo Fisher Scientific, Waltham, Massachusetts) were used for the TLD evaluation. The signal was integrated over the whole temperature range to avoid errors due to temperature differences of the individual TLDs and the reader.

The publication of Nunn *et al*¹⁰ shows that the measured TLD signal per unit air kerma depends on the effective energy, which is defined based on the half value layer (HVL) and homogeneity coefficient. Hence, the TLDs were calibrated with a Federal Institute of Metrology (METAS)-calibrated soft X-ray chamber (PTW type (PTW GmbH, Freiburg, Germany) 23342; 0.02 cm³) with an appropriate beam quality using the conventional treatment device Xstrahl 150 (Xstrahl Medical Solutions, Surrey, United Kingdom) at 120 kV with a first HVL of 3.64 mmAl compared with the first HVL of the CBCT device being 3.96 mmAl and an effective energy of about 49 keV, respectively.¹¹ For the

Table 1 Dosimetric specifications of the 3D Accuitomo 170 in cone beam CT mode⁹

FOV	40 × 40 mm ²	80 × 50 mm ²	140 × 100 mm ²
CTDI _{vol}	4.57 mGy	6.96 mGy	8.18 mGy
DLP	13.8 mGy cm	34.8 mGy cm	81.8 mGy cm
DAP	401 mGy cm ²	818 mGy cm ²	2160 mGy cm ²

3D, three dimensional; CTDI_{vol}, volume computed tomography dose index; DAP, dose area product; DLP, dose length product; FOV, field of view.

numerical data evaluation, the TLD readout was additionally corrected for background, individual sensitivity, reader-offset (reference measurements with strontium 90) and supralinearity.¹² The dose rate dependence, influence of measuring depth (energy spectrum), distance to central axis of the beam, self-absorption as well as directional dependences of TLD response were considered to be negligible.¹³ The calibration measurement had a standard deviation of $\pm 1.86\%$. The accuracy of the measurement chamber was about $\pm 2.8\%$. With the propagation of errors, the uncertainty of absolute dose calibration of the TLDs is assumed to be in the range of 3.4%.

A RT-Humanoid phantom (Humanoid-System, Torrance, CA) with 2.5 cm slices filled with two TLDs for each of the 71 measurement positions was used to measure organ doses. Only the upper 10 slices of the phantom were used, as for dental examinations no significant doses are expected in the lower parts.⁸ Since a large number of mean organ doses for various FOVs with different isocentre positions should be determined, a large number of measuring points and TLDs was needed. However, the position of the organs in the primary beam is irregular and the measurements in the proximity of the primary beam have shown great variability.

The TLD locations are depicted in Figure 1. They correspond to organs and tissues listed in ICRP103¹⁴: skin surface (10 locations), bone marrow, bone surface, brain (14 locations), salivary gland (eight locations), lymphatic nodes (eight locations), thyroid (four locations) and oesophagus (one location). The TLDs were also located within organs, which were not specially assigned in the ICRP103 such as mucous membrane of the naso-, oro- and hypopharynx (one location each), paranasal sinus (six locations), pituitary gland (one location) and ocular lens (two locations). The TLDs measuring the skin surface dose were attached on the surface of the phantom. The TLD positions as well as the estimated fraction of organ irradiation needed for the determination of organ doses were chosen as described in Dula *et al.*⁸ The fraction of organ irradiation for the oesophagus was estimated to be 10% and for the skin surface and lymphatic nodes an estimate of 5% was used corresponding to Ludlow *et al.*¹⁵ The data of a study from Poretti *et al.*¹⁶ were used to determine the dose contribution of single measurement points to both bone marrow and bone surface. The effective dose is expressed in μSv and was calculated according to: $E = \sum w_T w_R D_T$, where w_T is the tissue weighting factor

Figure 1 Locations of the thermoluminescent detectors within the RT-Humanoid phantom with corresponding organs.

for tissue T , w_R the radiation weighting factor and the D_T the absorbed dose in the tissue T of the organs at risk. For the calculation of effective dose, the radiation weighting factor $W_R = 1$ for X-rays and the ICRP 2007 tissue weighting factors w_T were used.¹⁴

For all CBCT measurements, exposure settings of 90 kV, 5 mA and full rotation were used as well as 80 kV and 2 mA for all scout views. A small FOV of $40 \times 40 \text{ mm}^2$ was selected to simulate a study of a group of molar and premolar teeth of the maxilla, a FOV of medium size of $80 \times 50 \text{ mm}^2$ was chosen for the examination of the whole maxilla and a large FOV of $140 \times 100 \text{ mm}^2$ was performed to scan the whole skull without skullcap. To be as realistic as possible, the size of the FOV as well as the field-dependent isocentre position was chosen to be similar to those used in daily clinical routine. All exposures were measured with the corresponding scout views and CBCT in the standard imaging mode with 17.5 s exposure time. For other imaging modalities (high resolution, high fidelity, high speed), the dose increases linearly, because there are no physical differences, besides the different exposure time. The dose values have to be multiplied by a factor of $f = 1.76$ for the high-resolution and the high-fidelity mode, whereas the factor amounts to $f = 0.6$ for the high-speed mode.⁹ The phantom was exposed four times. In order to include the setup uncertainties into these measurements, the phantom was repositioned between the individual exposures. Thus, the TLDs accumulate the dose from all four exposures. The dose in one position was defined as the average of the two TLDs. The whole measurement and evaluation procedure was performed two times for each FOV. The average of the measured dose values of the two sessions was used for the evaluation.

Monte Carlo dose calculation

The dose calculations with MC simulations were carried out using the EGSnrc MC transport code^{17,18} with the C++ class library of EGS++.¹⁹ The EGSnrc is a well-examined and utilized MC transport code system. The cut-off energy was set to 1 keV for the electron transport and 10 keV for the photon transport. The photo effect, Compton effect and Rayleigh scattering were considered for the photon interactions. The rotating photon source was modelled as a two-dimensional Gaussian intensity distribution with the specified nominal focal spot size values and a tabulated spectrum determined with SpekCalc.^{20–22} The influence of the exact nominal focal spot size and shape is negligible with respect to the resulting dose distribution.¹¹

In order to include the RT-Humanoid phantom into the MC simulation, a CT scan (Philips, Amsterdam, Netherlands Brilliance Big Bore CT, 16 slice) was performed. Hence, a voxelized phantom with dose scoring voxel of $2 \times 2 \times 2 \text{ mm}^3$ was obtained.

To connect the MC-calculated dose distribution D_{MC} in units of cGy/particle with the measured dose

D_{measured} in units of cGy/mAs, a normalization factor k_{norm} in units of particle/mAs had to be defined. This normalization factor was determined with the integral dose of the total TLD measurement for the specific FOVs.

$$k_{\text{norm}} \left[\frac{\# \text{particle}}{\text{mAs}} \right] = \frac{D_{\text{measured}} \left[\frac{\text{cGy}}{\text{mAs}} \right]}{D_{MC} \left[\frac{\text{cGy}}{\# \text{particle}} \right]}$$

Therewith, it was possible to compare the dose distributions with each other. All differences of the TLD measurement and the MC calculations were specified relative to the maximal dose value of the measurements specifically for each FOV.

The TLDs are calibrated to dose in water. Hence, the results of the MC simulation should be in units of dose to water as well. As the secondary low-energy electrons deposit their energy very locally, the differences in the stopping power ratios can be neglected in the MC simulation. Instead, the impact of the different mass attenuation coefficients should be considered. The ratio of the mass energy absorption coefficients of cortical bone relative to water amounts to about 5.5,²³ considering an effective energy of 50 keV. Hence, the TLD measurement points in the CT scan were replaced with the medium water. As the primary beam entrances the irradiation object from all directions, the assumption that this voxel replacement does not influence the total dose distribution remarkably could be justified.

Setup uncertainties lead to uncertainties of the location of the isocentre in the measurements, which in turn result in a mismatch between MC dose calculations and the measurements. Thus, the location of the isocentre was shifted in the MC calculations in order to match simulated with measured data. The required isocentre shift was determined with a random search around the primary estimated position.

Results

Thermoluminescent dosimeter measurement dose distribution

In Table 2, the measured organ doses are listed for the three FOVs 40×40 , 80×50 and $140 \times 100 \text{ mm}^2$. The mean organ doses, dose heterogeneity within the organ in units of standard deviation of different measurement points in the organ in (%) as well as the maximal and minimal dose measured are specified.

All organ dose values increase independently of the organ location with increasing FOV, such that the field size may be considered as dose optimization parameter. For all three FOVs, the highest absorbed doses were the averaged skin dose in the primary beam with 4.06, 6.49 and 8.72 mGy and the organ dose of the parotid gland with 2.25, 4.2 and 7.63 mGy for FOVs 40×40 , 80×50 and $140 \times 100 \text{ mm}^2$, respectively.

Table 2 Calculated effective doses and determined mean organ doses with standard deviations of the measurements as well as minimum and maximum organ dose values listed for the three different FOVs 40×40 , 80×50 and 140×100 mm²¹

Organ dose values			
Tube voltage:	90 kV	Imaging Mode:	Standard
Tube current:	5 mA	Rotation:	360°
	CBCT FOV: 40×40 Scout Image: 60×60	CBCT FOV: 80×50 Scout image: 100×100	CBCT FOV: 140×100 Scout image: 140×100
Organ	Mean organ dose [mGy] ± 1 standard deviation (Min; Max)		
Skin, averaged	4.06	6.49	8.72
Red bone marrow			
Skull	0.64	1.15	2.38
Body	0.11	0.20	0.42
Maxilla	3.45	5.37	5.75
Mandible	0.28	0.52	6.00
Bone surface			
Skull	2.95	5.36	11.03
Body	0.52	0.94	1.93
Thyroid	0.07 ± 0.01 (0.06; 0.07)	0.14 ± 0.01 (0.12; 0.15)	1.08 ± 0.06 (1.04; 1.20)
Parotid total	2.25 ± 0.78 (1.42; 3.59)	4.20 ± 0.61 (3.19; 4.71)	7.63 ± 0.58 (6.93; 8.74)
Parotid left	1.71	4.40	8.07
Parotid right	2.79	3.99	7.18
Remaining salivary gland	1.15 ± 1.44 (0.27; 3.64)	1.64 ± 1.79 (0.57; 4.74)	7.26 ± 0.69 (6.30; 7.98)
Mucous membrane	0.59 ± 0.45 (0.15; 1.21)	2.14 ± 1.43 (0.3; 3.79)	4.33 ± 1.63 (3.05; 6.63)
Paranasal sinus	1.57 ± 1.44 (0.21; 4.55)	3.87 ± 1.6 (0.64; 5.54)	4.60 ± 1.60 (1.17; 5.88)
Pituitary gland	0.19	0.61	1.34
Brain	0.12 ± 0.08 (0.04; 0.34)	0.44 ± 0.40 (0.11; 1.37)	1.05 ± 0.83 (0.25; 2.78)
Eye lens	0.17 ± 0.01 (0.16; 0.19)	0.64 ± 0.06 (0.58; 0.70)	0.99 ± 0.05 (0.94; 1.04)
	Effective dose [μ Sv]		
	48	97	250

Another relevant factor of the dose distribution is the localization of the isocentre for small FOVs.⁵ This fact cannot be seen from the TLD measurements in Table 2 as the influence of the choice of FOV and the isocentre position cannot be clearly separated. The standard deviation of the single measurement points in the organs decreased for most organs with increasing FOV, which shows that the dose distribution within a radiosensitive organ becomes more homogeneous with larger FOVs.

The determination of the mean organ doses was carried out by means of two complete independent measurements. The mean of the absolute local and global (normalized on maximum dose) difference for the determined organ doses and single TLD measurement

positions are specified in Table 3. These are indicators for the reproducibility of the measurements.

The effective dose was determined and resulted in 48, 97 and 250 μ Sv for the three FOVs 40×40 , 80×50 and 140×100 mm², respectively. In Figure 2, the dose averaged over each phantom slice is illustrated. For the individual measurement positions, the dose distributions in the phantom are indicated with a colour wash scale.

The increasing dose values with increasing FOV can also be seen in Figure 2. As expected, a dose contribution could also be measured outside the length of the reconstructed cylinder. This dose was not only due to scattered radiation, but also from the primary beam, since the shape of the beam is a cone. Therefore, the

Table 3 Mean absolute local and global difference of two independent measurements for organ doses and individual TLD measurement positions for the FOVs 40×40 , 80×50 and 140×100 mm²

FOV	40×40 mm ²	80×50 mm ²	140×100 mm ²
Mean local absolute difference of organ doses	6.2% (σ = 6.8%)	6.9% (σ = 5%)	3.4% (σ = 1.5%)
Mean global absolute difference of organ doses	1.8% (σ = 2%)	2.1% (σ = 3.5%)	1.6% (σ = 1.6%)
Mean local absolute difference of single TLD positions	6.3% (σ = 6%)	6.9% (σ = 9.5%)	4.2% (σ = 3.6%)
Mean global absolute difference of single TLD positions	0.9% (σ = 1.6%)	1.6% (σ = 4.2%)	1.5% (σ = 1.7%)

FOV, field of view; TLD, thermoluminescent dosimeter.

Figure 2 Mean dose for the single layers of the phantom with the corresponding radiosensitive organ or tissues within the layer of the phantom for different FOVs (top to bottom; 40×40 , 80×50 and 140×100 mm²) and the dose distribution for single TLD measurement positions.¹¹ FOV, field of view; TLD, thermoluminescent dosimeter.

Table 4 Calculated effective dose and statistical uncertainties of the MC calculations for the different FOVs of 40×40 , 80×50 and 140×100 mm²

FOV	40×40 mm ²	80×50 mm ²	140×100 mm ²
Effective dose	49 µSv	105 µSv	240 µSv
Maximal dose	38 mGy	46.4 mGy	65.5 mGy
Minimal dose	$7 \cdot 10^{-6}$ mGy	$3 \cdot 10^{-5}$ mGy	$1 \cdot 10^{-4}$ mGy
Maximal error	0.04 mGy	0.12 mGy	0.55 mGy
Average local D > 20% error	0.07%	0.3%	2.63%
D > 50%	0.04%	0.06%	0.11%
D > 90%	0.03%	0.08%	0.24%

FOV, field of view; MC, Monte Carlo.

cylinder volume can only be as high as the height of the beam at the surface of the phantom.

Monte Carlo dose distribution

In Table 4, the calculated effective dose and the statistical uncertainties of the MC simulations are shown for the different FOVs. The calculated effective dose amounts to 49, 105 and 240 µSv for the three FOVs 40×40 , 80×50 and 140×100 mm², respectively. For the statistical analysis, no voxels filled with air are taken into account, since these voxels are predominantly outside of the phantom. The determination of the average local uncertainties is based on all voxels receiving a dose >20, >50 and >90% of the maximal dose.

In Figure 3, the MC-calculated dose distributions are shown semi-transparently for the CT data set, which

has been used for the simulations. The dose planes were selected to specifically indicate the position of the isocentre. The scale was chosen to be logarithmic and linear, ranging from the minimum to the maximum dose of the three dose distributions.

In agreement with the TLD measurements, the calculated dose generally increases with the increasing size of the FOV. With the MC simulation, the resulting dose distributions can be determined and visualized in three dimensions. For the smallest FOV of 40×40 mm² the axial slice shows nicely the eccentric isocentre and hence also an asymmetric dose distribution in the xz-slices.

In Figure 4, the comparison between TLD measurement and MC simulation is provided for the results of single measurement positions as well as the global differences. The results are normalized to the integral of the all measurements. The double standard deviations of the global differences relative to the maximum value are ± 7.4 , ± 6.7 and $\pm 10.1\%$ for the FOVs 40×40 , 80×50 and 140×100 mm², respectively. The larger FOV proved to be much less sensitive to differences in the positioning of the patient as the smaller FOV.

When the dose values for the corresponding the radio-sensitive organs and tissues found with TLD measurements were compared to those, which had been determined with the MC simulation, a good correlation was found (Figure 5). The results are displayed for all three different FOVs 40×40 , 80×50 and 140×100 mm².

In Figure 6, the differences of the MC calculated dose values and the corresponding TLD measurements are shown as an average for the corresponding organs for the three FOVs 40×40 , 80×50 and $140 \times$

Figure 3 (a–c) Dose distribution of MC simulation for RT-Humanoid phantom for isocentre matching measurements in logarithmic (left) and linear (right) scale for the different FOVs of (a) 40×40 , (b) 80×50 and (c) 140×100 mm².¹¹ FOV, field of view; MC, Monte Carlo.

Figure 4 Comparison of the TLD measurement with the MC simulation for the three different FOVs (a) 40×40 , (b) 80×50 and (c) 140×100 mm². At the bottom are provided the global differences relative to the maximum dose value of the TLD measurement. FOV, field of view; MC, Monte Carlo; TLD, thermoluminescent dosimeter.

100 mm². They are specified relative to the maximum dose. The mean organ dose differences from the MC calculated dose values relative to the corresponding TLD measurements for all measuring points, amount to 0.4 ± 3.6 , 0.0 ± 3.8 and $-0.3 \pm 6.6\%$ (mean ± 2 SD) for the FOVs 40×40 , 80×50 and 140×100 mm², respectively.

Discussion

Factors influencing thermoluminescent dosimeter measurements

All mean organ doses had been determined with two independent measurement series within a RT-Humanoid anthropomorphic phantom. The mean local

Figure 5 Comparison between the determined dose values for the corresponding organs from TLD measurements with the MC simulation for the three different FOVs 40 × 40, 80 × 50 and 140 × 100 mm². FOV, field of view; MC, Monte Carlo; TLD, thermoluminescent dosimeter.

Figure 6 Differences between MC calculated dose values and the corresponding TLD measurements of organ doses relative to the TLD measurements for the three different FOVs 40×40 , 80×50 and 140×100 mm². Additionally, the double standard deviations for the individual measuring points within the organs are specified. FOV, field of view; MC, Monte Carlo; TLD, thermoluminescent dosimeter.

absolute differences for the determined mean organ doses between the two independent measurements are 6.2% ($\sigma = 6.8\%$), 6.9% ($\sigma = 5\%$) and 3.4% ($\sigma = 1.5\%$), while the mean local absolute differences for the single TLD measuring points amount to 6.3% ($\sigma = 6\%$), 6.9% ($\sigma = 9.5\%$) and 4.2% ($\sigma = 3.6\%$) for the FOVs 40×40 , 80×50 and 140×100 mm², respectively (Table 3). Obviously, a large absolute difference in mean organ dose determination corresponds to a large absolute difference of the single TLD measurements, whereby, as expected, a larger standard deviation in the measurement difference can be observed with the single position. These differences between series of organ dose determination can originate from changes of daily output of the CBCT device, positioning uncertainty as well as the uncertainty of the TLD measurement itself. No significant systematic error was found from the first to the second measurement series.

The mean differences between the two independent measurements and also its standard deviations are smallest for the largest FOV, partly due to the fact that there is more scattered radiation. Thus, the dose gradients at the field edges are reduced. Additionally, the field edges are at positions at the top and bottom of the phantom, where the density of measurement points is low. The measurements are therefore less sensitive to positioning errors for the large FOV compared with smaller FOVs. However, an attempt to reduce the influence of the positioning uncertainty was already made by repositioning the phantom between each of the four exposures during one measurement series.

With reduction of the size of the FOV, the exposed volume as well as the amount of scattered radiation decreased. Therefore, the mean dose can be reduced substantially as seen by the TLD measurements.

To determine the received dose when using the high-resolution or the high-fidelity mode based on the

measured dose values using the standard imaging mode, the dose values have to be multiplied by a factor of $f = 1.76$, whereas the factor amounts to $f = 0.6$ for the highspeed mode. These factors originate simply from the different exposure times used, as there is no physical difference between these different imaging modes.⁹ The imaging mode is therefore a simple parameter for dose optimization as the received dose scales linearly with the selected mode.

The choice of the measuring positions and the determination of the mean organ dose correspond to the study of Dula *et al*.⁸ In this study, local mean organ doses were measured from fan beam CT for a “standard” and two “low-dose” examination modes of the maxilla with the same RT-Humanoid phantom. The “standard” examination was defined as 40 slices at 165 mAs, and the “low-dose” examinations as 40 or 25 slices at 125 mAs, respectively. Compared with our results, the dose values from CBCT are substantially lower than the one for the fan beam CT. In the study of Dula *et al*.⁸ the equally determined mean organ dose values led to about 6.7, 5.2 and 3.1 mGy for the above-mentioned examinations. For the three FOVs 140×100 , 80×50 and 40×40 mm² of our study, the values amount to 5.2, 2.75 and 1.5 mGy, respectively, although two of the FOVs are larger. Consequently, CBCT examinations lead to a substantial dose reduction compared to conventional CT examinations.

Several previous studies^{3,5–7} performed TLD measurements to determine effective dose values from CBCT devices. In all these studies, the results were given as effective dose. Our effective dose range of 48–250 μ Sv for the small FOV of 40×40 mm² to the large FOV of 140×100 mm² is comparable to those dose values found by other studies such as 54–303 μ Sv,³ 19–368 μ Sv,⁵ 17–396 μ Sv⁶ for small to large FOVs. In the study of Ludlow *et al*.¹⁵ values of the effective dose for the same

machine settings of the 3D Accuitomo 170 device can be found. The effective dose for the three similar FOVs of 40×40 , 100×50 and 140×100 mm² with the 3D Accuitomo 170 were specified as 32, 113 and 269 μ Sv, which are within reasonable bounds. It is important to note that they found a remarkable variation in the CBCT dose for different technical settings and isocentre locations for the same CTBT unit. One example is the study of Morant *et al*²⁴ presenting organ dose as well as effective dose values for an i-CAT dental CBCT system. Effective dose values ranging from 25 to 66 μ Sv were lower as in the present study, however, a different CBCT system with different setups and protocols were used. In general, especially the size of the FOV can be seen as one of the most important factors for patient dose,^{3,5} which can be confirmed in our study. The mean organ dose values presented here are in the same order of magnitude compared to the result of the study of Silva *et al*.⁷

Comparison of calculated and measured dose distributions and factors influencing Monte Carlo dose calculation

As shown in Figure 4, a good agreement between the mean organ dose calculated with the MC method and the mean organ dose measured with TLDs in an anthropomorphic phantom was achieved. The double standard deviations of the global differences (relative to the maximum value) are ± 7.4 , ± 6.7 and $\pm 10.1\%$ for the FOVs 40×40 , 80×50 and 140×100 mm², respectively. A study of Morant *et al*²⁵ performed a validation of an MC dose calculation tool, however, designed for an i-CAT CBCT system. They achieved relative dose differences between TLD measurements and MC calculation of $\leq 13\%$ for measurement positions within the primary beam. The developed MC dose calculation tool was further applied to derive a conversion factor between measured dose area products and simulated effective dose values for several protocols, showing the broad applicability of device-specific MC dose calculation algorithms.²⁴ In the present study, the comparison between MC simulation and TLD measurement for the three different FOVs 40×40 , 80×50 and 140×100 mm² (Figure 5) showed no systematic error for the measuring points belonging to specific organs. A summary of this evaluation can be found in Figure 6. The organ-specific mean differences from the MC-calculated dose values relative to the corresponding TLD measurements amount to 0.4 ± 1.8 , 0.0 ± 1.9 and $-0.3 \pm 3.3\%$ for the FOVs 40×40 , 80×50 and 140×100 mm², respectively. Additionally, the relative differences of the calculated effective doses from the MC dose distribution and the TLD measurements are within $\pm 10\%$ for the three FOVs 40×40 , 80×50 and 140×100 mm², respectively.

All differences were evaluated relative to the maximum value of the TLD measurements. The dose distributions agree well with the TLD measurements. Measuring points located in regions with high-dose gradients might be more sensitive to positioning errors and therefore

might also show bigger differences compared with the MC simulation what is especially crucial for small FOVs. To obtain absolute dose values, the MC simulation was normalized with the integral dose of the TLD measurements. However, it would be beneficial to apply a single normalization factor, which may be determined with one single measurement for each FOV. Hence, a reproducible and well-defined reference situation needs to be found. If the implementation of the source in the MC simulation would further be improved, a single normalization factor for all FOVs might be possible.

Therefore, the accuracy of the technical specifications of the CBCT device is a natural limit for the MC simulation accuracy and needs to be considered as well. Parameters as the energy spectrum or field size can alter the resulting dose distribution remarkably.

As the MC calculation is independent besides the needed normalization factor, it would be possible to simulate also other protocols or CBCT devices by changing the technical parameters in the input. It exists a big variety of CBCT devices, which are different in many technical aspects, such as geometry, range of FOV, tube potential, filtering and others.

A CBCT device delivers a substantial higher dose than planar imaging,²⁶ which greatly motivates the development of an algorithm to accurately estimate the dose in the patient and consequently derive the risk to the patient. As the MC calculation results in a 3D dose distribution, the estimation of organ doses could be carried out more accurately than with single point dose values within specified organs. The MC dose calculation could therefore be a valuable tool to support the dentists in the evaluation of the trade-off between additional information, which may be relevant to the choice of therapy and the additional dose given to the patient.

Conclusion

The TLDs were calibrated specifically for the utilized energy and HVL of the CBCT. The error of the absolute dose calibration is assumed to be in the range of 3.4%. Reliable results of the mean organ dose values as well as effective dose values were achieved with TLD measurements in the RT-Humanoid phantom with a mean reproducibility within the range of 4.2–6.9% estimated based on the mean local absolute difference of single TLD positions for the range of FOVs from 40×40 to 140×100 mm², respectively.

The results of the MC-calculated organ dose values showed a good agreement with the TLD measurements. Considering the scope of diagnostic doses of a few mGy, the uncertainties are small enough to investigate, evaluate and compare dose distributions originating from different machine settings and patient positions with the developed MC simulation tool. Further studies regarding the normalization factor of the MC simulation need to be performed.

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