RESEARCH ARTICLE

Investigation of dental cone-beam CT pixel data and a modified method for conversion to Hounsfield unit (HU)

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Objectives: To investigate the relationship in dental cone-beam CT (CBCT) between the manufacturer-reported image pixel data and a modified conversion to CT number densities in Hounsfield unit (HU).

Methods: A standardized CT phantom was imaged using typical clinical parameters on CBCT from three manufacturers (Carestream 9300®, Carestream Health, Rochester, NY; J Morita 3D Accutomo®, J. Morita Mfg. Corp., Kyoto, Japan; and Planmeca Promax 3D®, Planmeca Helsinki, OY, Finland). Reconstructed axial slices were evaluated using regions of interest to ascertain the mean pixel value in five materials in the phantom. The Digital Imaging and Communications in Medicine data were also evaluated to determine if raw pixel data had been adjusted during the image reconstruction. A modified version of the existing manual HU conversion technique was applied, and the resultant slope and y-intercept were used to scale the pixel values ultimately to HU for all images.

Results: The DICOM header data show that a modified rescale y-intercept was applied to both the Carestream and Planmeca image data yielding manufacturer-produced results in HU. The Morita pixel data were unmodified and report in shades of grey or grey values (GV). The Carestream manufacturer-derived HU measurements showed good correlation in air (−1000 HU), but all other materials ranged from 2.6 to 13.5 σ from the specified phantom value. Results in the modified conversion technique images fell within 1.0–2.4 σ from the specified phantom values.

Conclusions: While more studies are needed to test for regularity, this study suggests that our modified technique could be a means of getting more accurate quantitative data from dental CBCTs.

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Introduction

Since the introduction of cone-beam CT (CBCT), this imaging technique has become extensively used in craniofacial diagnostic imaging due to its low cost, accessibility and reasonable radiation doses. An essential aspect of imaging techniques is the ability of the image to express density of an anatomic region being studied,

especially for hard and soft tissue pre-surgical structural assessment for procedures such as implant placement and identification of margins for jaw resections.

In conventional multiple-row detector CT (MDCT) the material density is quantified by the CT number, which describes the material attenuation properties in each voxel relative to the attenuation properties of water. Hounsfield unit (HU) is the value used to represent CT number and by definition has a value of 0 HU for water.

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MDCT scanners are calibrated by the manufacturer to water and air references for each available energy station and combination of beam filtration.¹ Additionally both the manufacturer and American College of Radiology (ACR) have recommended frequencies at which the users must perform quality control to ensure the MDCT system continues to report accurate CT number values for these reference materials. The ACR mandates that the value for water in the quality control phantom be 0 ± 7 HU.² On MDCT scanners these HU values are accurate and readily accessible without additional derivation by the user.

It is not unreasonable to expect dental CBCT to report values of tissue densities in conventional CT number as well. However, while MDCT's routinely provide relatively accurate material, or tissue, densities due to substantial manufacturer calibration and user quality control measures, a wide range of variability still exists when determining anatomic density values in dental CBCT including the manufacturer inconsistency between reporting voxel values in computed grey values (GV) vs HU. In a study by Katsumata *et al*,³ it was reported that an identical sample of bone produced variations of GV from -1500 to +3000.

Other studies have explored the use of CBCT evaluations of mandibular cortical bone as an alternative to the dual-energy X-ray absorptiometry (DEXA) scan with little success and highly variable readings.⁴ This variability is attributed to multiple factors, including the lack of standardization in tube potential, filtration and reconstruction algorithms of different manufacturers.⁵ Despite the inconsistency, a paper by Mah *et al*⁶ showed a strong correlation between the author-derived HU from CBCT GV and HU values obtained in the same material when imaged with conventional MDCT. Reeves and co-authors⁷ showed that there was a good linear relation between GV and attenuation coefficients, which was then used to calculate HU from measured GV. They concluded that linear regression from GV directly to HU at specified energies results in greater consistency. Stemming from such findings, this study aims to investigate the relationship in dental CBCT between the image pixel data reported by the manufacturer and development of an accurate conversion method to standardized CT number densities in HU based directly on the individual CBCT unit's beam energy and filtration used in the image acquisition.

Methods and materials

Radiographic phantom and imaging parameters

The standardized ACR CT 464 Phantom produced by Gammex (Middletown, WI) was selected for this study. The phantom contains four modules that correspond to various attributes evaluated on a standard MDCT scanner including CT number accuracy (Module 1), low contrast resolution (Module 2), CT number uniformity (Module 3) and spatial resolution (Module 4). Module

1 contains five cylinders comprised of different materials for assessing CT number accuracy in HU. The five materials, which represent a range of -1000 to 955 HU, are air, polyethylene, acrylic, and water and bone-equivalents as shown in Figure 1 along with a visual representation of the axis planes and example of the physical imaging set-up of the phantom in the CBCT.

Four commercially available dental CBCT units are used in clinical practices at our academic site. Of the four, the CS9300 (Carestream Health, Rochester, NY), 3D Accuitomo (J. Morita Mfg. Corp., Kyoto, Japan) and ProMax 3D Mid (Planmeca OY, Helsinki, Finland) were selected based on a scan field of view large enough to encompass the five materials in the phantom. The clinical imaging techniques used in this study are outlined in Table 1 with the corresponding CBCT manufacturer and model. In addition to phantom imaging, a calibrated solid-state radiation detector (RTI Black Piranha, Mölndal, Sweden) was used to measure the X-ray tube half value layer (HVL) for each CBCT unit at 90 kVp.

The phantom specifications for the materials in the CT number accuracy module are published by the ACR at 120 kVp.² Since CT numbers are energy dependent, specified values at 90 kVp were derived by scaling relative difference in the linear attenuation coefficients (μ) for each material at 120 kVp vs 90 kVp. These linear attenuation coefficients were determined using the chemical composition of each material with the National Institute of Standards and Technology (NIST) X-ray mass attenuation tables, and material densities.⁸ Table 2 shows the original ACR specification values at 120 kVp, and the derived specification values for 90 kVp. A study by Cropp *et al*⁹ investigated the change in CT number in this phantom for various energy stations ranging from 80 to 140 kVp on multiple MDCT scanner models and manufacturers. Since 90 kVp is not a common energy station for many MDCT scanners, the values for 80 and 100 kVp, which are also listed in Table 2, were selected to confirm that our derived HU specification values at 90 kVp are reasonable.

Reconstructed axial slices were evaluated using regions of interest (ROIs) to ascertain the mean pixel value in the five phantom materials. The Digital Imaging and Communications in Medicine (DICOM) header data for each image set were also accessed to determine if the manufacturer had adjusted the raw pixel data during the automated image reconstruction process. The phantom images and DICOM header data were viewed using Radiant DICOM Viewer version 3.4.2 (Medixant, Poznan, Poland).

CT number conversion

The X-ray tube voltage during each CBCT is recorded as the peak voltage (*i.e.* 90 kVp); however, this is not the average energy of the primary X-ray beam. Mah *et al*⁶, suggested that the best correlation value (R^2) of multiple linear regressions fit to data across a range of energy

Figure 1 Phantom configuration and experimental setup. (a) The CT number accuracy section (module 1) of the CT ACR 464 phantom contains five cylinders composed of different materials. (b) The phantom was positioned on a stand in the gantry of the CS9300 to acquire images. Similar set-ups were used with the other two CBCT models. (c) An example of a reconstructed axial slice from module 1. (d) A graphical representation of the three-dimensional planes of the phantom as specified in the study. CBCT, cone-beam CT; ACR, American College of Radiology.

levels up to the peak voltage would correspond to the appropriate average energy of the beam.

Similarly, we used the NIST table data to ascertain linear attenuation coefficients for the five materials contained in the ACR phantom. Unlike the method by Mah *et al*⁶, the measured HVL for each X-ray tube was used as a determinate for average beam energy rather than the best R^2 value. The HVL is an easily measured value representing the thickness of a material, which reduces the intensity of an X-ray beam entering that material by one-half. The HVL of a material is inversely related to the linear attenuation coefficient of a given material, which in this case is aluminium (Equation 1).

$$\text{HVL} = \frac{\ln(2)}{\mu_{\text{Aluminium}}} \quad (1)$$

Using the measured HVL in millimeters of aluminium, the linear attenuation coefficient of aluminium ($\mu_{\text{Aluminium}}$) was calculated for each CBCT X-ray tube at the same peak voltage as the selected clinical parameters. The energy corresponding to the calculated linear attenuation coefficient in the NIST data for aluminium was established as the average beam energy.

Table 1 Dental CBCT imaging equipment and selected clinical protocol parameters

CBCT Model	Tube voltage (kVp)	Tube current time product (mAs)	Matrix	Pixel size (μm)	Slice thickness (mm)
CS9300	90	45	557 × 557	300	0.3
3D Accuitomo (A3D)	90	123	512 × 512	250	1.0
ProMax 3D Mid (PM3D)	90	109	500 × 500	400	0.4

CBCT, cone-beam CT.

Table 2 CT number values for the five materials in the CT Phantom at various kVp stations

Peak energy station (kVp)	Air	Polyethylene	Acrylic	Water-equivalent	Bone-equivalent
80 ^a	−980 ± 13	−128 ± 9	106 ± 7	1 ± 2	1154 ± 193
90	−1000	−106	112	0	1091
100 ^a	−983 ± 12	−106 ± 5	120 ± 6	1 ± 3	1004 ± 110
120 ^b	−1000	−95	120	0	955

The ACR material values in the CT phantom are specified at 120 kVp, and the 90 kVp values are derived from these. The values listed at 80 kVp and 100 kVp are average values with standard deviation from a published study.

ACR, American College of Radiology.

^aCropper *et al* averages with standard deviation.

^bACR recommended values.

At the extrapolated average energy for each image, linear attenuation coefficients were derived from the NIST data for each of the five phantom materials. A linear regression was fit to this data to obtain a theoretical slope and y-intercept for converting mean pixel values in GV to linear attenuation coefficient values. The resulting linear attenuation coefficient (μ) values were converted to CT number (HU) using Equation 2 for each image pixel.

$$HU_{\text{material}} = 1000 \times \frac{\mu_{\text{material}} - \mu_{\text{water}}}{\mu_{\text{water}}} \quad (2)$$

The image data were imported to MATLAB (The Mathworks Inc, Natick, MA) to scale individual pixel data as described above. This resulted in post-processed images with pixel values in HU that could be viewed in any DICOM viewing software. For image datasets with rescale y-intercepts already applied to the pixel data, the data were scaled back to GV and then re-processed using this modified conversion method.

Results

CBCT & DICOM header data

Direct HVL measurements at 90 kVp for the three CBCT systems were relatively consistent with an average of 6.71 ± 0.61 mm aluminium. The smallest HVL was measured on the CS9300 at 6.17 mm, the median was the 3D Accuitomo (A3D) at 6.58 mm and the largest was the ProMax 3D Mid (PM3D) at 7.37 mm. Table 3 shows the derived average energy values from the NIST tables for each system at a peak imaging energy of 90 kVp. The average energies ranged from 46.7 to 51.7 keV and were approximately 54% of the peak voltage setting.

The DICOM header data for each image set show that a modified rescale y-intercept of -1000 is applied during the automated reconstruction process in both the CS9300 and PM3D image data yielding manufacturer-reported pixel values in HU. The DICOM header data for the A3D show the reconstruction of the pixel data is unmodified and reporting in GV.

Table 3 Measured HVL, derived average beam energy (E_{AVG}) and average energy vs peak X-ray tube voltage during exposure expressed as a percentage

CBCT	HVL (mm Al)	E_{AVG} (keV)	Rescale		
			E_{AVG}/kVp	y-Intercept	Slope
CS9300	6.17	46.7	52%	0.0560	0.0006
A3D	6.58	48.4	54%	0.2582	0.0007
PM3D	7.37	51.7	57%	-0.1018	0.0004

CBCT, cone-beam CT; GV, grey values; HVL, half value layer. Rescale y-intercept and slope for conversion of pixel GV to linear attenuation coefficients (μ) are also listed for each CBCT.

DICOM header data also showed varying slice thickness between manufacturers for the chosen clinical protocols, which ranged from 0.3 to 1.0 mm as shown in Table 1. To evaluate a comparable volume in module 1 of the phantom for all three image sets a number of reconstructed axial slices were selected based on this variation in slice thickness. Therefore, 5 slices for the A3D, 12 slices for PM3D and 16 slices in the CS9300 image sets were evaluated.

For each image set, ROIs were drawn in the five phantom materials in consecutive axial slices. Figure 2 shows the variation in material measurement for each image set along the z-axis. The ROI measurements are reported in this figure as reported by the manufacturer in either HU or GV as applicable. The largest z-axis deviation was seen in the A3D images for bone with a standard deviation (σ) of 10.6 GV. The remaining materials showed ≤ 1.5 GV with the exception of air, which showed a variation of 6.8 GV. Similarly, the PM3D images showed the largest z-axis variation in bone (4.4 HU) as well. The CS9300 varied ≤ 5.7 HU for all materials. Interestingly the air value showed no variation along the z-axis, and all air ROIs reported -1000 HU with no standard deviation in pixel values within each individual ROI in the CS9300.

CT number

The CS9300 and PM3D both reported pixel values in HU. The average values for each material ROI in consecutive z-axis slices with one standard deviation are listed in Table 4 with the subscript Male to signify the manufacturer-reported HU. For each ROI along the z-axis the standard deviation within the ROI of pixel values was also recorded and the average standard deviation is listed in Table 3. Using the DICOM specified rescale y-intercept value reported by these two manufacturers the pixel data were converted back to GV. The GV pixel data for all three manufacturers were converted using the modified method described previously to pixel values in HU. As a result of modifying each individual pixel throughout the imaging volume in MATLAB for all three image sets, the z-axis averages of the ROI pixel data as described above are listed in Table 4 as well with the subscript Mod.

Comparison of both the manufacturer-reported (Male) and modified conversion (Mod) HU values were made to the derived 90 kVp specifications for the ACR phantom in each material. Table 5 shows the number of standard deviations necessary from both the manufacturer-reported and modified HU values to ensure the derived specification value for each material falls inside the uncertainty range. Notably σ was unable to be measured for the CS9300 air value. The reported air value is -1000 HU and the modified conversion value is -803 HU, thus showing no difference from the derived air specification value to the manufacturer-reported HU and only a 197 HU difference for the modified air value. All material HU values for the modified technique fell within 2.4σ yielding a 96% confidence level of the specified material HU value. This

Figure 2 Variation in pixel values along the z-axis for each material from the image sets acquired on the (a) CS9300, (b) PM3D and (c) A3D. $\square$, polyethylene; $\circ$, water; $\diamond$, acrylic; $\blacktriangle$, bone; $\blacktriangledown$, air.

was not true for the manufacturer-reported HU values, which ranged from 2.6 to 13.5 σ .

Discussion

This study demonstrates a relationship between the raw pixel values of CBCT and the converted values in HU.

Table 4 ACR CT phantom module 1 CT number values

CT number (HU)					
CBCT	Air	Polyethylene	Acrylic	Water-equivalent	Bone-equivalent
CS9300Man	-1000	-822 $\pm$ 79	-679 $\pm$ 81	-710 $\pm$ 78	-67 $\pm$ 86
PM3DMan	-783 $\pm$ 31	-190 $\pm$ 32	-41 $\pm$ 33	-117 $\pm$ 32	469 $\pm$ 48
CS9300Mod	-803	-372 $\pm$ 194	-1 $\pm$ 196	2 $\pm$ 189	1494 $\pm$ 210
A3DMod	-1129 $\pm$ 54	-159 $\pm$ 61	84 $\pm$ 64	1 $\pm$ 59	953 $\pm$ 69
PM3DMod	-1080 $\pm$ 51	-117 $\pm$ 51	128 $\pm$ 54	4 $\pm$ 54	957 $\pm$ 81

ACR, American College of Radiology; CBCT, cone-beam CT; HU, Hounsfield units.

All the converted values in HU of the ACR Phantom were within 2.4 σ when compared with the known HU values of the phantom, showing a clear resemblance to the CT number values expected by MDCT. In contrast, the manufacturer-reported HU values had a far greater range from 2.6 to 13.5 σ . As illustrated in the study by Katsumata *et al*³ and in this study, the manufacturer HU derivation method is not only inconsistent but also clinically unreliable. A new method, such as the one

Table 5 The necessary number of standard deviations of the average CT number for the derived specification values at 90 kVp for each material to be within the uncertainty range

CBCT	Air	Polyethylene	Acrylic	Water-equivalent	Bone-equivalent
CS9300Man	-	9.0	9.8	9.1	13.5
PM3DMan	7.1	2.6	4.7	3.7	12.9
CS9300Mod	-	1.4	1.0	1.0	1.9
A3DMod	2.4	1.0	1.0	1.0	2.0
PM3DMod	1.6	1.0	1.0	1.0	1.7

CBCT, cone-beam CT.

investigated in this study, for deriving HU values is a necessity for accurately demonstrating the densities of any anatomic regions on CBCT scanners.

Interestingly, the CS9300_{Man} air value and DICOM header data suggest a truncation of the raw pixel data to an artificial y-intercept point prior to the HU conversion. When the CS9300_{Man} pixel data are converted back to greyscale using the DICOM reported rescale values, all air values are always scaled to zero, effectively calibrating the image data to air. This suggests that the manufacturer's reconstruction process is designed to shift the darkest GV to zero prior to applying the manufacturer-produced rescale algorithm for conversion to HU. This results ultimately in a loss of pixel data and an unknown truncation value, which may explain the large CS9300_{Mod} average standard deviation values (194–210 HU). It is also the reason we chose not to report average standard deviation values for air in the CS9300_{Man} and CS9300_{Mod} data.

Although this is not the first study to propose that raw pixel values of CBCT can be converted into HU values (with relevant clinical implications) by evaluating the average X-ray energy, this study improves upon previous methodologies by utilizing the HVL for each individual X-ray tube to determine an average X-ray energy specific to each scanner. On the other hand, the study by Mah and colleagues⁶ derived the average effective energy from theoretical R^2 value, which was not scanner specific. Furthermore, while using multiple scanners, they were able to produce some linear results for certain materials, but the accuracy of the water value, which is crucial in determining the validity of all other values, were inconsistent and widely variable. This is most likely due to their non-scanner specific method for determining the linear attenuation coefficients at the determined average energy. In contrast, one of the most significant results from our study is that the modified HU conversion method was able to produce the water value of 0 HU within 1.0σ for all three CBCT scanners because the linear attenuation coefficients from the determined average effective energy are specific to each CBCT X-ray tube. As all HU values are by definition relative to the water density levels, which is represented numerically as 0 HU, the accuracy and consistency of the water value is crucial in determining the validity and success of any studies involving HU values. Moreover, assessing the application of CBCT HU values in clinical settings requires the scanner to be calibrated in relation to water. In other words, without confirming that the water HU value is 0, all other HU values become meaningless.

These findings have the potential to be applicable in numerous clinical settings. An accurate diagnostic imaging and density assessment in bone is crucial in implant surgery because implant failures rate tends to be higher when placed in poor quality bone.^{10,11} Having a CBCT volume that can provide correct bone density values will allow the implant surgeon to better

determine prognosis of implants while increasing the survival rate. Furthermore, having such pre-operatively determined bone density information means that the implant surgeon can rely less on experience to subjectively assess the quality of bone. Pre-operative planning with reliable hard tissue imaging is an essential part of oral and maxillofacial surgery treatment modalities as well. For example, accurate mandibular radiographic imaging of bony metastasized oral squamous cell carcinoma is crucial to visualize the extent of bone for resection. Studies performed previously on this topic show promising marginal advantages of CBCT over conventional CT in these kinds of cases.¹² Another good clinical implication in accurate CBCT density assessment is the evaluation of Medication Related Osteonecrosis of the Jaw (MRONJ). Stage 1 and 2 MRONJ lesions currently are indicated for surgical debridement while more severe Stage 3 MRONJ lesions are indicated for resections.¹³ Block resections of MRONJ are currently recommended due to inability to identify borders of the MRONJ lesion. These resections ensure the complete removal of all necrotic and compromised tissue to prevent possibility of reestablishment of the lesion. Yet, this is achieved by removing significant amount of healthy and uncompromised bone surrounding the MRONJ lesion leaving the patient with extensive co-morbidities and disfiguration.¹⁴ The ability to accurately and precisely identify and demarcate the borders of MRONJ lesions by analysis of bone density, in conjunction with other determining factors such as pertinent clinical data and the visual radiographic texture of the bone, would potentially shift the recommended surgical treatment from non-conservative block resections to effective surgical debridement and conservative resections.¹⁵ Hence having an accurate pixel density value is everything when identifying the areas that are necrotic or malignant. In the preimplant imaging, accurate bone density value determines the success rate of the implant structure.

The modified HU conversion method presented in this work has a few inherent limitations. The conversion method is modified to derive average beam energy based on the HVL measurement at the peak energy station. If the imaging protocol is changed to different peak energy stations, the HVL and resulting average beam energy would also change. Furthermore, the HVL is also specific to the X-ray tube and its inherent filtration from the surrounding tube housing. This varies between X-ray tube manufacturers and models. Thus, the modified method presented in this work would be applicable to other CBCTs, but the actual average beam energies and rescale values are not directly applicable to other clinical CBCTs. Additional studies are also needed to investigate the validity of this method when applied to images with unknown standard CT number densities such as patient images.

Additionally other sources of error also need to be considered and solutions must be sought. CBCT images

are subject to artefacts deriving from a variety of sources including: presence of metallic artefacts, X-ray hardening, data discontinuity-related artefact in subject objects larger than the CBCT field of view.^{16,17} These confounding effectors disturb the possibility of CBCT use for accurate density analysis.

Conclusion

In summary, the study was able to formulate a valid conversion technique that can provide HU readings for three CBCT units. The HU values from CBCT, derived from raw pixel values (GV), were comparable to those of MDCT within only 2.4 σ . Most importantly, consistent water values within 1.0 σ throughout all three CBCT scanners was the pivotal result, which indicates a need to further test this method in a clinical setting. Testing may include conducting a retrospective study comparing

patients who may have had both CT and CBCT scans and applying the conversion technique used in this study. This would be a true clinical assessment of the accuracy of this method.

For CBCT equipment that report pixel, or voxel, values in HU, such as two of the CBCTs used in this study, users must be extremely cautious when analysing this data for clinical patient care. As was shown in this study, the manufacturer-reported HU values can be significantly unreliable showing a wide range of variability in accuracy ranging from 2.6 to 13.5 σ for various materials depending on the manufacturer's calculation of HU. While further clinical studies are needed to test for regularity, based on research evidence, an equation, such as the method used in this study, could be a pathway to obtain more accurate and reliable data that are clinically meaningful rather than relying on inconsistent HU values displayed by the CBCT scanner.

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