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Detection and Evaluation of Specimen-Mass Changes with a 16-bit Intraoral Imaging Charge-Couple Device

Hassem Geha

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DETECTION AND EVALUATION OF SPECIMEN-MASS CHANGES
WITH A 16-BIT INTRAORAL IMAGING CHARGE-COUPLED DEVICE

Hassem Geha
DDS, Saint Joseph University, 1997
DUA, Lebanese University, 1999
DUB, Lebanese University, 2001

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DETECTION AND EVALUATION OF SPECIMEN-MASS CHANGES
WITH A 16-BIT INTRAORAL IMAGING CHARGE-COUPLED DEVICE

Presented by

Hassem Geha, DDS, DUA, DUB

Major Advisor

Alan G. Lurie, DDS, PhD

Associate Advisor

Michael L. Freedman, PhD

Associate Advisor

Sanjay M. Mallya, BDS, PhD

Associate Advisor

Charles F. Hildebolt, DDS, PhD

Associate Advisor

Lamont R. MacNeil, DDS, MS

University of Connecticut

2005
DEDICATIONS

To Gracia, with all my Love.

To Lilia, Theodore, Grace and Sara,

To Marie-Helene and Georges,

To Samih and Rindala, Samah and Lama, Sameh and Zeina, Raghed and Maya,

To Mom and Dad,

To Assine and Elias,

To Mountaha, Marmar and Uncle Simon,

To Adib, Nazira, Hicham and Rayan
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I. INTRODUCTION

Since Roentgen’s discovery of X-rays in 1895, film has been the primary medium for capturing, displaying, and storing radiographic images. This technology is the most common and convenient for dental practitioners. In the 1980’s, there was advancement in medical and dental imaging with the introduction of digital radiography, and this technology is slowly replacing conventional, film-based radiography.

Digital imaging can be divided into two categories: direct and indirect digital imaging. Direct digital imaging incorporates computer technology in the capture, display, enhancement, transmission and storage of radiographic images. Such digital images are acquired with charge-coupled devices (CCD), complementary metal oxide semiconductors (CMOS) or photostimulable storage phosphor (PSP) plates. Indirect digital imaging involves digitization of conventional films or dental radiographs with a camera or scanner. The introduction of direct and indirect digital imaging systems in medicine and dentistry has revolutionized diagnostic imaging by improving diagnostic quality, automating image analysis, enhancing treatment-planning and improving patient education.

The process of direct or indirect digitization of images involves the conversion of continuous, uncountable data, to well-defined, finite and countable data. Assuming an object is composed of a continuum of elements, its analog image will contain the same data quantity and require the same countless number of elements to represent it. To process such an image by a computer, it must first be converted to a digital form of
discrete elements called pixels. This conversion applies to all attributes that form the image. In the case of direct digital imaging, the attributes are the data-acquisition modality (CCD, CMOS, or PSP) and exposure settings. But in the case of indirect digital imaging, the factors included are the analog-to-digital converter (scanner or camera) and the factors that are involved in the formation of the image on a conventional film: exposure settings and film processing.

Several clinical studies used x-ray film as the “reference standard examination” and compared the diagnostic performances of digital systems and film systems for caries and periapical lesion detection. Dove and McDavid showed that there was no significant difference in diagnostic accuracy between non-enhanced digital images and conventional film-based images for the detection of proximal-surface caries. A study by Uprichard, however, found that CCD-based direct digital radiography was not as accurate as conventional film images in detecting inter-proximal caries in mixed dentition, but suggested that, with increased experience, direct digital images could be as accurate as film for diagnosis. Wallace showed that conventional films outperformed digital images in their diagnostic efficacy for periapical lesions. The major advantage of direct digital imaging was its lower radiation dose.

Another method to evaluate and compare the diagnostic performance of radiographic systems is the perceptibility-curve test which is a method for evaluating the ability of radiographic imaging systems to record and display minute changes in absorber densities. This test is used also to evaluate observers perceptions of density changes with
conventional radiographic systems, direct digital radiographic systems (CCD and PSP), and indirect digital radiography. The results suggest that if digital systems are used properly, and the images are enhanced digitally, the perception of density changes on digital radiographs can exceed the perception of density changes on film-based radiographs\textsuperscript{7,8,14-18}

A limiting factor in diagnostic radiography is the human visual system that is able to distinguish only about 60 shades of grey under optimal viewing conditions\textsuperscript{19}. In conventional radiography the visual human perception of shades of grey is based on differences in the optical densities of the film, but in digital radiography, the perception of shades of grey is based on pixel grey values, which are translated into luminance on a computer monitor. Digital image acquisition is based on 8-bit (256 shades of grey), 10-bit (1024), 12-bit (4096), 14-bit (16384) or 16-bit (65536) data, whereas the computer monitor display is 8-bit (256 shades of grey), but 8-bit computer monitors do not display more than 242 grey values\textsuperscript{19}. Moreover, Chen et al found that there was a 2nd degree polynomial (i.e. a non-linear) relationship between pixel grey values and display screen luminance\textsuperscript{20}. With these three limitations, useful information, therefore, is lost between acquisition and display even with digital image enhancement, and diagnostic accuracy can be compromised.

An advantage of digital imaging over conventional film-based imaging is that one can do a quantitative analysis of digital images. Several methods have been developed for this purpose: histogram analysis and bone mass measurement.
Radiometric differentiation of radicular cysts and granulomas, with histogram analysis, was evaluated in two studies\textsuperscript{2, 4}. Both studies digitized, with a camera, conventional radiographs of known cysts and granulomas and compared the gray level histograms of the lesions. There was disagreement between the two studies about whether cysts and granulomas could be differentiated radiometrically. The study's use of an indirect digital imaging rather than direct digital system might have accounted for these results.

Couture and Hildebolt\textsuperscript{21}, using a new PSP system, presented a quantitative model for bone mass measurements in digital oral radiography. They stated that adequate quantitative measurement of bone mass from an intraoral radiograph depends on accurate measurement of x-ray attenuation, and that detection of small changes in bone mass depends on highly reproducible images and consequently on great instrumental precision. These measurements were possible because of the linear response of the PSP to increasing exposure. Because they were interested in estimating the lower limits of detection of small changes in bone mineral content, they included in this model corrections for soft tissue, scattered radiation, and changes in exposure.

One disadvantage of using PSP for data analysis is that images acquired with PSP systems are subject to degradation if time elapses between exposure and processing of latent images\textsuperscript{21, 22}. Images from CCDs are not subject to such degradation since the latent image data is immediately processed by the computer\textsuperscript{23}. Moreover, the mathematical
relation between x-ray exposure and the final digitized output of the CCD is linear within a range of exposures\(^{17}\). With this characteristic of the CCD, I hypothesize that the CCD used in this work will permit quantitative measurement of changes of mass of an absorber.

One of the most common diagnostic challenges in oral radiology is the detection of subtle pathological changes\(^{20}\). By analyzing the grey values changes from a CCD across an ROI, one might detect small lesions in mineralized dento-maxillofacial tissues before they become visible on either a digital display or on film. The objective of this study is, therefore, to test the hypothesis that the mass of intraoral structures (i.e. bone, teeth) can be calculated by evaluating numerically and graphically the changes in their grey values obtained with a 16-bit charge-coupled device (CCD).

To obtain accurate quantitative measurements, I divided the work into four consecutive experimental steps. In step A, I evaluated the output of the x-ray machine for its consistency and linearity, and I determined the relationship between the changes of exposure with increasing thickness of an aluminum absorber. In step B, I evaluated the reproducibility and linearity of the CCD sensor to the x-ray exposure. In step C, and after validation of the equipment in steps A and B, I determined the relationship between changes of grey values on the CCD sensor and the increasing thickness of standardized test object (aluminum stepwedge), and generated a mathematical model for this relationship. In step D, I applied the mathematical model to grey values of a digital
radiograph of an intraoral biological specimen and related the changes of these grey values to corresponding changes in thicknesses of standardized test objects.
II. MATERIAL AND METHODS

A CCD chip is an array of radiation-sensitive elements, which are small, silicon-based semi-conductors that form the sensitive area of the chip. Each x-ray photon that reaches the CCD displaces some electrons creating electron-hole pairs in the depletion region of the silicon. The amount of electron-hole pairs created is proportional to the number of photons. The depletion region has a maximum capacity for holding electron-hole pairs; the condition when this maximum capacity is reached is called saturation and it is caused by overexposure of the CCD. These electron-hole pairs are localized in small delimited areas called pixels. The pixels are arrayed in lines and are gated to allow the transfer of one pixel line to the next. The last line of the array transfers the electron-hole pair into a horizontal-shift register called a read-out (Figure 1). This horizontal-shift register allows the transfer of one pixel to the next (in the manner of a fire-brigade), and the last pixel of this horizontal register, is connected to an output gate. The output gate of the CCD array is connected to an analogue-to-digital converter to convert the electrical signal into a digital picture, which is stored in computer memory.

The x-ray unit used was a Focus intraoral x-ray machine (Instrumentarium Imaging Inc., Tuusula, Finland) operating at 70 kVp, 7 mA. Exposure times were varied between 0.02s and 0.25s. With a focal spot of 0.7mm and added filtration of 2 mm Al, the radiation field was 60 mm in diameter at 22.9 cm from the focal spot. The x-ray exposure and absorbed exposure were measured with a dosimeter (Radiation Monitor Controller, Model 2026C, Radcal Corp). The CCD was a 16-bit, size 2, intraoral sensor (Cygnus Ray
MIS, Cygnus Technologies Inc., Scottsdale, AZ, 85260, USA). The pixel size is 22µm and the signal to noise ratio is 45:1. The manufacturer claimed that the optimal performance of the CCD encompasses exposure times from 0.04s to 0.08s, and that the edges of the image obtained with the CCD will appear slightly “brighter” on the monitor screen because of the adhesive used to couple the silicon chip to the electronics in the CCD. The manufacturer also recommended, for data measurements, an interval of 4 to 5 minutes between exposures to allow cooling of the CCD, thus reducing the effect of dark current noise. This time interval is not required in day-to-day dental radiography [personal communication with Tony Bavuso from Cygnus Imaging].

The test objects and absorbers used in these studies were (1) 39 sheets of light aluminum, each 0.3mm thick, (2) Stepwedge 1 with 13 steps of aluminum, (3) stepwedge 2 with 7 steps of aluminum, and (4) one 6.4mm thickness of Lucite®, employed as soft tissue-equivalent absorber in some experiments. The steps of stepwedges 1 and 2 were measured with a vernier with a precision of 0.05mm. A human dry mandible with teeth was also used. It had random holes of 0.5mm, 1mm and 1.5mm in diameter and depth on its lingual aspect. A jig was used (Figure 2) to obtain reproducible positioning of the x-ray machine, ionization chamber, CCD receptors, and test objects. In all experiments, the distance from the x-ray source to the ionization chamber and to the CCD receptor was 40 cm.

For image acquisition, Cygnus Imaging provided customized software (Sensor Communication v.1.0.16) to enable access to the raw data and to acquire and save the
images as 16-bit TIFF images. For evaluation of grey values, I used NIH Imaging Software (ImageJ 1.32). Microsoft® Excel® was used for data analysis, graphic plots, equation generations and calculation of $R^2$ for best fit equations. Best fits and $R^2$ were generated from Microsoft® Excel® Trendline algorithm. $R^2$ was used to evaluate the goodness of the fit.

The results were evaluated based on their clinical significance. Clinical significance is a term that will be used in this the discussion of the results of this work. The definition of "clinical significance" is difficult to tackle. Clinical significance is generally used to imply that a small effect has a clinically beneficial or harmful value. A small change in exposure that does not have any effect on the visual diagnostic efficacy of the image, but that significantly changes the data registered on the CCD is considered to be clinically significant. The 16-bit sensor records 65536 grey values. Considering the noise sources on the CCD image (i.e. photon noise, dark noise) a fluctuation of grey values of more than 5% (3260 grey values) on the plate was considered clinically significant. When comparing mean grey values from two images acquired at the same exposure settings, a difference between mean grey values more than 500 grey values was considered clinically significant. Bohay stated that 30% to 60% of mineral loss is required to produce radiographic evidence of disease\textsuperscript{24}, which is often referred to as clinically significant, thus, if the difference between the calculated aluminum thicknesses and the actual aluminum thickness is more than 5%, this difference was considered clinically significant.
A. Glossary

**Aluminum Equivalent Thickness (AET):** A known thickness of aluminum that attenuates the x-ray beam to the same degree as a mass of tissue, e.g. bone.

**Blank exposure:** Exposure taken with no object between x-ray source and receptor.

**Consistency:** Reproducibility of data (exposure output, grey values) with the ionization chamber and the CCD receptor.

**Dark current:** Electron-hole pairs that are not read-out and remain in the detector, over time, independent of photons incident on the CCD detector. Dark current arises from thermal energy within the silicon lattice of the CCD of these residual electron-hole pairs.

**Histogram:** Distribution of grey values within an ROI. With a 16-bit CCD, the grey values vary between 0 (black) and 65535 (white).

**Photon noise:** Statistical fluctuation of x-ray photons absorbed by the CCD.

**Saturation:** Condition where the depletion region of the silicon in a CCD has reached its full capacity of holding electron-hole pairs.

**Transfer function:** Mathematical relationship between x-ray input and digital output of a CCD.

**Uniformity of CCD:** The variation in grey values within an image is clinically not significant.
III. RESULTS

A. Step A: Evaluation of X-ray unit

In the following experiments, the x-ray receptor used was the ionization chamber. The purpose of these experiments is to test the consistency and the linearity of the x-ray tube, and to evaluate the changes in measured exposure relative to the increasing thicknesses of aluminum absorber.

1. X-ray tube output consistency

This experiment was done in triplicate. In experiments 1 and 2, the exposure time was 0.25s and in experiment 3 it was 0.16s. In experiment 1 and 2, 15 consecutive exposures were taken, and the output was measured in mR (mRoentgens). There was a slight increase of x-ray output from one exposure to another, but after 8 exposures the output became consistent (Figure 3). Similar observations were seen in experiment 3.

2. X-ray output linearity

For this experiment (and for all the experiments that followed), the x-ray machine was warmed up with 8 consecutive exposures at 0.25s, and the exposures were measured in mR at the following exposure times: 0.04s, 0.06s, 0.08s, 0.10s, 0.12s, 0.16s, 0.20s and 0.25s, using one exposure at each setting. Figure 4 shows a linear x-ray tube output in the range of exposures used with $R^2=0.999$. 
3. Change of exposure with increasing aluminum absorber thicknesses

In this experiment, the exposure time was set at 0.25s. The 0.3 mm aluminum sheets were placed between the x-ray source and the ionization chamber in increments of 2 sheets. The measured exposure was plotted against the increasing aluminum thickness. The best fit between the measured exposure in mR and increasing absorber thickness was a 5th degree polynomial equation with a value of $R^2=0.9999$. Figure 5 shows the plot of the relation on a normal scale, and Figure 6 shows the plot on a semi-logarithmic scale.

B. Step B: Evaluation of the CCD

In the following experiments, the receptor was the 16-bit Cygnus size 2 CCD. The purpose of these experiments is to evaluate the uniformity of the CCD response to x-ray exposures and to evaluate the linearity of the CCD response to increasing exposure times.

1. Uniformity of the CCD response

Twenty blank exposures, with no aluminum absorber between x-ray source and CCD receptor, were made at various exposure times between 0.02s and 0.10s. On the computer monitor all the exposures produces similar images and the images acquired at higher exposure times were darker than the ones taken at lower exposure times. Each image contains 2069680 pixels arranged in 1640 rows and 1262 columns. On all the images, the first 12 columns on the left edge of the image and the first and last rows had
no data on them (Figure 7); both therefore were excluded from the analysis. The grey values histograms from all images had a normal distribution shape (Figure 8).

A digital radiograph acquired at 0.04s was arbitrary selected; its mean grey value was 46004. The mean grey values for the columns were evaluated. Figure 9 shows that there is a decrease of mean grey values from 50000 to 46000 for the first 50 columns after which the mean grey values fluctuated between 45800 and 46200, and for the last 50 columns the mean grey values increased to 47000. Similar results were observed when the mean grey values of the rows were evaluated. The edges of the image had higher mean grey values than the center (difference between grey values more than 500 grey values). The same analysis was done on the same image after removing the first and last 50 columns and rows, and the mean grey values fluctuated between 45840 and 46131 and the difference between the mean grey values was less than 500 (Figure 10, Table 1). The fluctuation of the grey values at 0.04s, before removing the edges, was more than more than 3260 grey values (Figure 11a) and after removing the edges the fluctuation was less than 3260 grey values (Figure 11b). The edges of the CCD image, therefore, were excluded from all subsequent analysis and the remaining grey values on the image were considered uniform.

2. Reproducibility of the CCD response

In the following two experiments, exposures were made from 0.02s to 0.16s, and ROIs of 800x800 pixels were selected in the center of the image.
This experiment was made in 2 stages. In the first stage the x-ray tube was not warmed up with 8 exposures at 0.25s and 25 to 30 exposures were made at each time setting. In the second stage, the tube was warmed up and 15 to 20 exposures were made at each time setting. The mean grey values of the ROIs of images acquired at the same exposure setting were compared. When the tube was not warmed up there was a decrease in the mean grey values and the difference between the mean grey values was more than 500 grey values (Figure 12). When the x-ray tube was warmed up, the difference between the mean grey values of the ROIs was less than 500 grey values (Figure 13), and the CCD was considered to have a consistent response.

3. Linearity of the CCD response

The parameters established in the previous experiments, in which the response of the CCD and the output of x-ray generator were consistent, were used. When the mean values of grey from each setting were plotted against the increase in exposure time, there was a linear response of the CCD between 0.02s and 0.12s ($R^2=0.9987$), after which the linearity was lost (Figure 14).

C. Step C: CCD response to increasing aluminum absorber thickness

The purpose of the following experiments is first to find a relationship between increasing thickness of aluminum and changes of grey values, second, to validate this
relationship, and third to evaluate whether or not adding soft tissue equivalent thickness changes the relationship, and how.

1. Change of mean grey values with changes of absorber thickness

In the following experiment, the 0.3-mm aluminum sheets were added one-by-one between the x-ray source and the CCD against the CCD. The exposure time was set at 0.04s. ROIs of 800x800 pixels were selected in the center of the image, and the mean grey value from each image was plotted against the increasing thickness of aluminum. The best fit relation was a 5th degree polynomial equation with a value $R^2=1$ (Figure 15).

2. Verification of the 5th degree polynomial fit

Stepwedge 1 and stepwedge 2 were placed, consecutively on the CCD and exposed at 0.04s. From the image of each step of stepwedge 1, I selected an ROI of 100x300 pixels, and calculated its mean grey value. When the change of mean grey values was plotted against the increasing thickness of the steps, the fit was a 5th degree polynomial (Figure 16, E1) with a value of $R^2=1$.

From the image of stepwedge 2, the mean grey value of an ROI of 100x300 pixels from the image of each step was extracted. From the equation E1 generated previously with stepwedge 1, I calculated the thickness of aluminum that corresponded to the mean grey values of stepwedge 2. This calculation was done with the Solver algorithm from
Microsoft® Excel®; a customized-Solver algorithm was written for the purposes of this work. Table 2 shows the difference between the actual thickness of stepwedge 2 and the corresponding calculated thickness. The difference between the calculated and actual thicknesses of aluminum was less than 5%.

3. Adding soft tissue equivalent

This experiment tests whether adding a soft tissue equivalent (Lucite®) changes the shape or the goodness of the 5th degree polynomial fit. I added 6.4mm of Lucite® over stepwedge 1 and used an exposure time of 0.04s. The mean grey values of the ROI’s were extracted as in the previous experiment, and the grey values were plotted against increasing thickness of the stepwedge. The fit was 5th degree polynomial (Figure 16).

4. How adding the soft tissue equivalent changed the coefficients of the fit

The purpose of this experiment is to assign an aluminum equivalent thickness (AET) for the soft tissue equivalent. Stepwedge 1 was used as a reference for the AET of the Lucite®.

The Lucite® thickness was added between the x-ray source and stepwedge 2 against the stepwedge and parallel to the CCD. The exposure time was 0.04s. The mean grey values corresponding to each step of the stepwedge were measured. For each step, “S”, of stepwedge 2, the AET of Lucite® plus aluminum was calculated from equation E1 (AET_s). Subtracting the aluminum thickness “S” from AET_s yielded the AET of the
Lucite® alone (AET_L). Table 3 shows the calculated values of AET_S and AET_L for each step, “S”. Figure 17 shows a linear relation between AET_S and “S” with R²=1. In figure 18, the relation between AET_L and “S” was approximated by a linear curve with R²=0.8841.

D. Step D: Clinical application

The previous experiments showed a specific relationship between grey values and increasing thickness of absorber. In the following experiment, to test the clinical applicability of this relationship, I analyzed the changes in grey values on a digital image of a dry mandible and, with equation E1, I calculated the AET that corresponded to these grey values changes. The ICRU report 4425 presents tables of mass attenuation coefficients of tissue equivalents. The mass attenuation coefficients of aluminum and cortical bone are almost the same, when exposed to photon energies in the range of dental x-ray machines (Figure 19)25. Thus, an AET can be assigned to the bone.

1. Apply findings to dry mandible

In this experiment, the molar region of a dry mandible was exposed for 0.04s with no addition of soft tissue equivalent. A large ROI selected for analysis (Figure 20). Within this ROI, I averaged the mean grey values from non-overlapping small ROI’s of 10x10 pixels (0.2x0.2mm). Using equation E1, derived from the aluminum step wedge 1, I calculated the AET for the corresponding mean grey values.
Figure 21 shows the results of a 10x10 pixel scan of the ROI in an intraoral molar projection, plotted with a 3-dimensional graph. The graph shows the changes of AET of bone (z-axis) and the width (x-axis) and height (y axis) of the ROI. This figure shows the shape of a mandibular molar, a pulp chamber, and, distally, an extraction socket. The labeled colors are the assigned values of AET, and the change in colors indicates an increase or decrease in the AET. The plot shows the presence of one of the prepared holes (a) apical to the roots of the molar where the AET of the bone varies between 2.5mm and 4mm. The area of increased density in the missing tooth site (b) represents the denser part of the lamina dura in this region. Mesial to the middle third of the molar mesial root there is a small area of decreased AET (c) with a minimal AET value of 2.5mm to 3mm. On the radiograph, this corresponds to a small radiolucency that looks like a trabeculation, but is, in fact, a 1mm-hole drilled on the lingual aspect of the dry mandible in this area.
IV. DISCUSSION

The major limitation of the x-ray machine is that it needs a number of exposures to warm-up before producing consistent output (Figure 3). The manufacturer states that the x-ray machine needs several “warm-up” exposures before the output becomes consistent and that the output of the tube is slightly lower when the unit is cold [personal communication with Ari Jarvinen from Instrumentarium Imaging]. When the x-ray tube was not warmed up, the error in the difference between the mean grey values acquired at the same exposure setting was clinically significant (Figure 12). When the x-ray tube was warmed up, the error was not clinically significant (Figure 13). The limitation of the CCD receptor is that it needs a four-to-five minute interval between exposures to allow it to cool. In their physical evaluation of a 10-bit Dixel® CCD receptor, Yoshiura et al. showed that there was no effect of dark current\textsuperscript{17}. They did not specify the time between their exposures, although they did state that acquiring the images at 10-bits and storing them at 8-bits might have reduced the effect of dark current noise\textsuperscript{17}. Yoshioka et al. reported a decrease in pixel grey values in RVG-S CCD images and related it to dark current. He also stated that the effect of dark current noise can be corrected so that it will not interfere with the formed image\textsuperscript{26}. This might lead to a dramatic reduction in wait-time between clinical exposures.

For all experiments, the x-ray machine output was consistent after the warm-up exposures (Figure 3), and the response of the Cygnus 16-bit CCD was also consistent after the warm up exposures (Figures 12, 13). Because of this, reproducible
measurements were possible from one exposure to another. This consistency conforms to the model proposed by Couture and Hildebolt\textsuperscript{21}, who stated that to obtain accurate quantitative measurements of mass changes, there should be accurate measurements of x-ray attenuation and highly reproducible images.

Chen et al.\textsuperscript{27} evaluated a RVG 3200 ZHR intraoral CCD detector response to x-ray exposure and found that the central part of the detector had lower pixel values than the peripheral areas. In another study\textsuperscript{20}, the same authors showed that the response of the CCD is heterogeneous with relatively low ability to reflect changes in mass of imaged test objects in the central part of the CCD. My study demonstrated that the edges of the image had higher grey values than those of the central part, with the central part being relatively large (1500x1150 pixels, Figures 9 and 10), and that the fluctuation of the grey values and the difference between the mean grey values were less than the error set as being clinically significant (Figure 11b, Table 1).

Yoshiura et al. showed a slight non-linear response of the 10-bit Dixel\textsuperscript{®} CCD sensor to increasing dose\textsuperscript{17}, but they considered it to be linear nonetheless. I found a linear response over a range of exposures after which the linearity was lost (Figure 14). This is most likely due to the physical properties of a CCD: at high exposures, the well capacity of a pixel is exceeded, causing a saturation of the well\textsuperscript{17}.

The best fit of the response of the ionization chamber and Cygnus 16-bit CCD to increasing aluminum thickness was a 5th degree polynomial function (Figures 5, 6, 15
and 16). Other functions were tried (2nd, 3rd and 4th degree polynomial with respective $R^2$ values of 0.98, 0.999 and 0.9995, and the exponential fit had an $R^2$ value of 0.98). The difference between the actual and calculated thicknesses of the aluminum stepwedge 2 with the 2nd, 3rd and 4th polynomial equations and from the exponential equation was more than 5%. Therefore, these equations were not used in this work. A possible explanation for the fit not to be exponential with $R^2=1$ is the filtration and hardening of the x-ray polyenergetic beam.

In the verification of the 5th degree polynomial equation E1, I used relatively large ROI's (>10000 pixels/ROI). This might explain the clinically non-significant differences between the actual and calculated aluminum thicknesses values in Table 2. Couture and Hildebolt21 stated that when they decreased the size of their ROI, they had lower accuracy in their measurements compared with large ROI's. In my experiment on the dry mandible, I scanned the image with only one size ROI (10x10 pixels, 0.2x0.2mm). If I were to have chosen a large ROI, I would have lost details by averaging together too many pixel grey values. If I were to have chosen an ROI of one pixel, I would have increased the noise in the results. Further investigation on the minimal ROI size is warranted in another study.

The Lucite® did not change the degree of the polynomial fit (Figure 16). Although the AET of Lucite® plus aluminum varied linearly with the increasing thickness of aluminum (Figure 17), there was a fairly linear relation ($R^2=0.8841$) between the AET of Lucite® and the increasing thickness of the aluminum stepwedge (Figure 18,
The results for figure 18 are counter intuitive in that as the step wedge gets thicker, the beam is hardened; so the x-rays pass more readily through it (that is, fewer are absorbed), and the thickness of Lucite would appear to be less, but the results suggest opposite. This might be due to the difference between the mass attenuation of Lucite® and aluminum in the range of energies used.

The three-dimensional plot (Figure 21) is a topographic representation of the AETs of a selected clinical film’s ROI (Figure 20). The radiographic anatomy can be related to the topographic view for interpretation. The enhanced digital radiographic image (Figure 20) shows a relative radiolucency in the apical region of the mandibular molar, but it gives no information about the extent of this radiolucency. When the 5th degree polynomial equation generated from the aluminum step wedge was applied to the grey values of the clinical film in this apical region, the results showed a loss of bone mass equivalent to 1.5 mm of aluminum thickness. Thus, the change of AET in designated areas of a clinical digital image indicate differences in bone mass, but, at this point, one cannot specify the buccal/lingual location of features responsible for these differences, nor if changes in these differences occur over time.
V. CONCLUSION

This study showed that there is a 5th degree polynomial relationship between grey values on the Cygnus 16-bit CCD and the increasing thickness of aluminum. This relationship can be applied to the grey values from an image of an intraoral specimen. The changes of bone mass, represented as changes in aluminum thicknesses, were evaluated numerically and graphically and the detection of these changes appears to be clinically significant.

In this study, I used one x-ray unit, one CCD, two stepwedges and a single dry mandible; therefore, the results can not be generalized on all equipment. The results, however, demonstrate that quantitatively evaluating changes of grey values might lead to detection of small changes in mass. This has considerable importance in radiographic diagnosis in that the ability to detect small changes in mass is critical to the early detection of caries, alveolar bone loss, and apical radiolucencies. This could lead to the improvement of diagnosis and treatment planning for these conditions and the improvement of the treatment outcome as well as the reduction of the costs and risks associated with these conditions. This would also be a step towards computerized diagnosis.
VI. FIGURES

Figure 1: Schematic of CCD

Figure 2: Setup of x-ray machine and receptor (ionization chamber or CCD) in all experiments
Figure 3: X-ray machine exposures for consecutive identical exposure settings

Figure 4: Change of exposure with increased exposure time (Equation and $R^2$ Value)
Figure 5: Measured exposure with increasing aluminum thickness (normal plot) (Equation and $R^2$ value)

$y = -0.001759x^5 + 0.060315x^4 - 0.8235x^3 + 5.9249x^2 - 26.118x + 75.283$

$R^2 = 0.9999$

Figure 6: Measured exposure with increasing aluminum thickness (Semi-logarithmic plot)

$y = \log_10 \left( -0.001759x^5 + 0.060315x^4 - 0.8235x^3 + 5.9249x^2 - 26.118x + 75.283 \right)$

$R^2 = 0.9999$
Figure 7: Blank digital radiograph showing columns and rows that are excluded from the analysis
Figure 8: Histogram of grey values from a blank exposure of a CCD at 0.03s
Figure 9: Change of mean grey values along the columns of the image at 0.04s

Figure 10: Change of mean grey values along the columns of the image after disregarding the data from first and last 50 columns at 0.04s
Figure 11: Histogram of grey values of a radiographic image acquired at 0.04s before removal of the first and last 50 columns and rows (a) and after removal of the first and last 50 columns and rows (b)
Figure 12: CCD grey values following consecutive identical exposure settings at 0.03s without warm-up exposures

Figure 13: CCD grey values following consecutive identical exposure settings at 0.04s after warm-up exposures
Figure 14: Change in grey values with increased exposure time

\[ y = -344487x + 59534 \text{ (between 0.02s and 0.12s)} \]
\[ R^2 = 0.9987 \]

Figure 15: Change in grey values with increasing aluminum thickness (equation and \( R^2 \) value)

\[ y = 0.11857x^5 - 4.2667x^4 + 63.935x^3 - 549.89x^2 + 3309.2x + 46071 \]
\[ R^2 = 1 \]
Figure 16: Change in grey values with increasing aluminum thickness. Blue curve: without Lucite®. Red Curve: with Lucite®. (Equations and $R^2$ value)
Figure 17: Change of AET of Lucite® + Aluminum (AET$_2$) relative to increasing thickness of aluminum (Equation and $R^2$ value)

$y = 1.0561x + 0.4413$

$R^2 = 1$

Figure 18: Change of AET of Lucite® (AET$_1$) relative to increasing thickness of aluminum (Equation and $R^2$ value)

$y = 0.0692x + 0.5934$

$R^2 = 0.8841$
Figure 19: Mass attenuation coefficients of cortical bone (blue) and aluminum (purple). The source of this figure is the ICRU-44 mass attenuation coefficients tables of cortical bone and aluminum.

Figure 20: Photograph and Radiograph showing the selected area of interest. ‘a’, ‘b’ and ‘c’ are the areas described in the text.
Figure 21: Three-dimensional plot of the molar region. The x- and y-axis represent the ROI width and height respectively. The z-axis represents AET. The labeled colors designate the change of AET. Each small square on the plot represents the 10x10 pixel ROI. “a”, “b” and “c” correspond to “a”, “b” and “c” in figure 20.
### VII. TABLES

<table>
<thead>
<tr>
<th>Minimum “Mean” Grey Value</th>
<th>Mean Grey value</th>
<th>Maximum “mean” grey value</th>
</tr>
</thead>
<tbody>
<tr>
<td>45840</td>
<td>46004</td>
<td>46131</td>
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</table>

Table 1: Fluctuation of mean grey values within the digital image with 0.04s exposure time and after disregarding the first and last 50 columns and rows

<table>
<thead>
<tr>
<th>Mean Grey Value</th>
<th>Actual Thickness (mm)</th>
<th>Calculated Thickness (mm)</th>
<th>Difference (mm) / Error %</th>
</tr>
</thead>
<tbody>
<tr>
<td>45757</td>
<td></td>
<td>0.00</td>
<td>0.00 / 0%</td>
</tr>
<tr>
<td>49741</td>
<td>0.95</td>
<td>0.94</td>
<td>0.01 / 1.05%</td>
</tr>
<tr>
<td>52586</td>
<td>2</td>
<td>2.04</td>
<td>0.04 / 2%</td>
</tr>
<tr>
<td>53803</td>
<td>2.75</td>
<td>2.75</td>
<td>0.00 / 0%</td>
</tr>
<tr>
<td>54975</td>
<td>3.7</td>
<td>3.69</td>
<td>0.01 / 0.27%</td>
</tr>
<tr>
<td>55905</td>
<td>4.65</td>
<td>4.70</td>
<td>0.05 / 1.07%</td>
</tr>
<tr>
<td>56726</td>
<td>5.85</td>
<td>5.81</td>
<td>0.04 / 0.68%</td>
</tr>
<tr>
<td>57297</td>
<td>6.7</td>
<td>6.76</td>
<td>0.06 / 0.89%</td>
</tr>
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</table>

Table 2: Comparison of calculated and actual aluminum thicknesses of stepwedge 2 relative to their corresponding grey values
<table>
<thead>
<tr>
<th>Step Thickness mm S</th>
<th>Grey Values</th>
<th>AET (Lucite® + Al) mm: AET&lt;sub&gt;S&lt;/sub&gt;</th>
<th>AET (Lucite®) mm: AET&lt;sub&gt;L&lt;/sub&gt;</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>48268</td>
<td>0.46</td>
<td>0.46</td>
</tr>
<tr>
<td>0.95</td>
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<td>1.43</td>
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<td>2</td>
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<td>57087</td>
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<td>6.7</td>
<td>57541</td>
<td>7.53</td>
<td>0.83</td>
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</table>

Table 3: Evaluation of AET of Lucite® + aluminum (AET<sub>S</sub>) and AET of Lucite® alone (AET<sub>L</sub>) relative to the increasing thickness of stepwedge 2
VIII. BIBLIOGRAPHY


DETECTION AND EVALUATION OF SPECIMEN-MASS CHANGES
WITH A 16-BIT INTRAORAL IMAGING CHARGE-COUPLED DEVICE

Hassem Geha
DDS, Saint Joseph University, 1997
DUA, Lebanese University, 1999
DUB, Lebanese University, 2001

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DETECTION AND EVALUATION OF SPECIMEN-MASS CHANGES
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Presented by

Hassem Geha, DDS, DUA, DUB

Major Advisor

Alan G. Lurie, DDS, PhD

Associate advisor

Michael L. Freedman, PhD

Associate Advisor

Sanjay M. Mallya, BDS, PhD

Associate Advisor

Charles F. Hildebolt, DDS, PhD

Associate Advisor

Lamont R. MacNeil, DDS, MS

University of Connecticut

2005
DEDICATIONS

To Gracia, with all my Love.

To Lilia, Theodore, Grace and Sara,

To Marie-Helene and Georges,

To Samih and Rindala, Samah and Lama, Sameh and Zeina, Raghed and Maya,

To Mom and Dad,

To Assine and Elias,

To Mountaha, Marmar and Uncle Simon,

To Adib, Nazira, Hicham and Rayan
ACKNOWLEDGMENTS

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Georges Allam, for writing the software algorithm that I used in this work.
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