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Digital image ratio : a new radiographic method for quantifying changes in alveolar bone. Part I: Theory and methodology.

A. Jean⁺, Y. Epelboin⁺⁺, A. Rimsky⁺⁺, A. Soyer⁺⁺ and J.P. Ouhayoun⁺

⁺Département de Parodontologie, Faculté de Chirurgie Dentaire, Université Paris VII, 5 rue Garancière, 75006 Paris

⁺⁺Laboratoire de Minéralogie-Cristallographie, UA 009 CNRS, Universités P.M. Curie et Paris VII, Case 115, 4 place Jussieu, 75252 Paris Cedex 05, France

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Address for reprints : Y. Epelboin.

Laboratoire de Minéralogie-Cristallographie, UA 009 CNRS, Universités P.M. Curie et Paris VII, Case 115, 4 place Jussieu, 75252 Paris Cedex 05, France

ABSTRACT

A new method, digital image ratio (DIR), has been developed for directly measuring changes in alveolar bone. The image on the computer monitor represents the relative mass change between two radiographs. A Fourier filtering is used to reduce noise artefacts. This method is validated through an experiment with a step wedge. DIR needs only a preliminary calibration of the experimental conditions of operation and avoids tedious calibrations for each measurement as in the case of Digital Image Subtraction. It is suggested to use low-voltage X-ray techniques for long-term quantitative studies of patients to minimise irradiation doses.

Theory and its validation are presented in a first part. A full clinical study appears in a second one.

Key words: Image filtering, image analysis, image ratio, bone radiography, quantitative digital radiography, quantification of bone changes.

INTRODUCTION

There are two accurate methods for visualizing and measuring the changes in bone mass using two superimposable radiographs taken at two different times. The first, digital subtraction radiography (DSR), has improved the diagnostic accuracy of dental radiology (1-8). The density of grey in the region of interest is converted into an equivalent thickness of the hydroxyapatite or aluminium standard. The density of the subtraction image can also be analysed by computer (CADIA), and the changes in the density of the alveolar bone are measured directly from the image (9). The second photodensitometric technique (10-11), requires a combination of photodensitometry and a computer to transform optical densities into aluminium-equivalent millimeters. However, neither technique is very convenient, since both require accurate calibration for each study. Moreover such conversions may introduce errors attributable to the heel effect and thus may not be warranted.

This study describes a new method based on the computation of the ratio of digitized radiographs. The resulting image, which is related to the mass ratio, directly shows the changes in bone mass. It is illustrated by the analysis of a few cases presented in a second part.

The main features of the method are shown in Figure 1.

THEORY

The analysis of a radiograph is the result of the following processes: exposure of the dental film, digitization of the radiographic image, measurement of the density of grey and calculation of the mass of the material whose changes are to be studied. In this section these different points are discussed.

Film exposure

The probability of a photon interacting with the silver bromide or iodide grains in the photographic emulsion depends on the energy of the photons, i.e. on the wavelength of the beam. For each wavelength in the incident beam, we can write (12,18):

$$D - f = D_{\max} (1 - \exp(-kX)) \quad (1)$$

where X is the illumination (i.e. the number of photons) impinging on the film at a given wavelength, k a characteristic of the photographic emulsion, $D_{\max}$ the maximum density capable of being recorded with the emulsion, i.e. a physical characteristic of the film. Its value varies between 4 and 6, depending on the emulsion. D is the optical density of the film and $\exp$ is the exponential function. The minimum density is f , called the fog level, which value is of the order of 0.1-0.2. Figure 2 shows a typical curve: Figure 2a shows D versus X , Figure 2b D versus $\text{Log}_n X$, which is the conventional plot. A plot of D versus $\text{Log}_n X$ (Fig. 2b) shows an area between D_a and D_b where the relationship between D and $\text{Log}_n X$ is linear:

$$D = \gamma \text{Log}_n X + D_0 \quad (2)$$

where D_0 and γ are constants. In all the calculations Log_n means the natural logarithm. The parameters f , $D_{\max}$ and γ depend on the physical characteristics of the film and on its processing (13). Standard working conditions must be defined.

Digitizing the radiograph

The optical density of grey, D , is defined as:

$$D = \text{Log}_n \frac{I_0}{I} \quad (3)$$

where I_0 is the intensity of the visible light impinging on the radiograph and I the intensity of the beam passing through. I , intensity of the visible light used when digitizing the radiograph, must not be confused with X , the number of photons absorbed when recording the radiograph. The camera records the number of visible photons

impinging on its detector. The value of a pixel is related to the intensity I thus a pixel value, in the digitized image, is directly related to the density of grey in the original radiograph (14).

Digitizing introduces noise, which limits the density of information, i.e. the number of levels which are distinguishable. It is always less than the maximum pixel value (14) and can be increased by summing videoframes (15). Noise artefacts are removed by filtering the images by means of Fourier transforms (14, 17).

Measuring the absorbing mass in a radiograph

Relationship between photoelectric absorption and absorbing mass: If X_i is the illumination (number of X-ray photons) incident on an object and X is the illumination that passes through the object, then :

$$X = X_i \exp \left(-\frac{\mu}{\rho} m \right) \quad (4)$$

where μ/ρ is the mass absorption coefficient of the material for the considered wavelength and m the total mass per unit of area lighted by the X-ray beam. μ/ρ depends on the material (16) (it is not the same for soft tissues, hydroxyapatite or calcium).

Equation (4) may be written as:

$$m = \frac{\rho}{\mu} \text{Log}_n \frac{X_i}{X} \quad (5)$$

When digitizing a radiograph, the intensity I of the visible light going through the lens of the camera is recorded in the memory of the computer. According to equations (3) and (2) if the density of grey corresponds to the linear part of the densitometric curve:

$$\text{Log}_n I = \text{Log}_n I_0 - D$$

and

$$D = \gamma \text{Log}_n X + D_0$$

thus $\text{Log}_n I = \text{Log}_n I_0 - D_0 - \gamma \text{Log}_n X$ (6)

From equations (5) and (6), it can be shown that:

$$m = \frac{\rho}{\mu} \left(\text{Log}_n X_i + \frac{1}{\gamma} \text{Log}_n I + \frac{D_0}{\gamma} - \frac{\text{Log}_n I_0}{\gamma} \right) = \frac{\rho}{\mu} \frac{1}{\gamma} (\text{Log}_n \alpha + \text{Log}_n I) \quad (7)$$

where $\text{Log}_n \alpha = \gamma \text{Log}_n X_i - \text{Log}_n I_0 + D_0$

is a constant for all radiographs recorded with the same experimental conditions for the X-ray machine and digitised in the same conditions.

This equation is written for a single wavelength; we assume for the sake of simplicity that the constants do not vary significantly for the range of wavelengths produced by the X-ray machine.

The ratio technique: Two different cases must be considered. When the densities of grey correspond to the linear part of the densitometric curve (i.e. between 1.6 and 3 about as measured by Kodak for Ultraspeed and Ektaspeed films) equation (7) is valid. For lower values one must use equation (5) only.

Let us consider first equation (7) for two different radiographs, at points corresponding to the same location. I_1 is the intensity passing through a given point in the first radiograph, I_2 the intensity at the corresponding point in the second one.

$$m_1 = \frac{\rho}{\mu} \frac{1}{\gamma} (\text{Log}_n \alpha_1 + \text{Log}_n I_1) \quad m_2 = \frac{\rho}{\mu} \frac{1}{\gamma} (\text{Log}_n \alpha_2 + \text{Log}_n I_2) \quad (8)$$

We have measured that the value of a pixel p in the memory of the computer is proportional to the logarithm of the intensity I impinging on the corresponding pixel of the camera CCD. Using a known optical standart (Zeiss Jena, Germany) which density varies from 0.1 up to 3.0 by steps of 0.1 we have plotted the pixel values of its digitized image versus the density value and we fund that:

$$p = a \text{Log}_n I + b \quad (9)$$

where a and b are two positive constants. This must be checked for other cameras and other experimental conditions. Thus equations (8) may be rewritten, replacing the intensity values I by the pixels values as:

$$\frac{m_2}{m_1} = \frac{(a \log_n \alpha_2 - b) + p_2}{(a \log_n \alpha_1 - b) + p_1} = \frac{C_2 + p_2}{C_1 + p_1} \quad (10)$$

Instead of computing the difference between the pixel values as in Digital Subtraction Radiography (DSR), we may use equation (10), which describes the ratio of the digitised images. This allows us to measure directly any changes in absorbing mass between two radiographs. The two-dimensional array m_2/m_1 for all pixels is called the “mass ratio image”. It is an array of real numbers which can be normalised between 0 and 255 to be displayed in black and white or with pseudo-colors on the screen of a computer.

Let us now consider the case when the densities of grey are too small so that equation (2) is not valid. This occurs for densities lower than 1.6. From equations (3) and (9), one may write:

$$D = \log_n I_0 + \frac{b - p}{a} \quad (11)$$

The constants are measured using a step wedge as explained later, in the validation section, thus the relationship between p and D is known. The relationship between X , the X-Ray illumination and D may be tabulated from experimental measurements. In our case the curve has been provided by Kodak for the operating and development conditions used for the clinical studies. Using equation (5), equations (8) are replaced by:

$$m_1 = \frac{\rho}{\mu} [\log_n X_i - \log_n X_1] \quad m_2 = \frac{\rho}{\mu} [\log_n X'_i - \log_n X_2]$$

which means:
$$\frac{m_2}{m_1} = \frac{\log_n X'_i - \log_n X_2}{\log_n X_i - \log_n X_1} = \frac{K_2 - \log_n X_2}{K_1 - \log_n X_1} \quad (12)$$

Equation (12) is equivalent to equation (10) except that the relationship between the pixel value p and the illumination X is no more straightforward. Both K_1 and K_2 as well as C_2 and C_1 are constants which should be the same if the operating conditions remain unchanged between both radiographs. If not contrast adjustment may be necessary (appendix A). Other methods have been suggested by different authors (19, 24-26). The determination of these constants will be explained in the next section.

When computing the ratio between pixels values the noise is additive and thus enhanced in the resulting calculation. High frequency noise is removed by applying a low-pass filter in the Fourier space.

An advantage of digital image ratio (DIR) is that it is equally valid for both soft tissues and bone, since the absorption coefficient does not appear in equations (10) and (12).

Validation

To determine the constants C or K in equations (10) or (12) and to validate the method we use a step wedge (build in the laboratory) made of 10 areas. Each area is made of superposed layers of metal of equal thickness, thus the mass, in each area, is proportional to the number of layers. The step wedge is exposed to the X-Rays and the radiograph is developed in the same experimental conditions as used for the patients. The radiograph is then digitised twice, producing two images. The second image is shifted so that the same location in both images corresponds to different thicknesses. We study such areas to compute their mass ratio. Since the thicknesses, thus the masses are known this experiment allows to calculate C and to validate the method.

First it is necessary to verify equation (9) and to measure the constants appearing in equation (11). As explained before, this is done digitizing a known optical standard. It is then easy to measure precisely the optical densities for any radiograph. Such a calibration is made only once if the camera is always operated in the same conditions.

We have found that the measured densities for the radiograph vary from 0.2 which is too low to consider equation (10) thus we had to use equation (12). This means

that X was determined from the measurement of the density D by means of the illumination curves provided by Kodak. This is computed automatically: the curves are registered in the computer which transforms the pixels values into optical densities using equation (11) then into $\text{Log}_n X$ using the tabulated curves.

Equation (12) may be written as:

$$K = \frac{\frac{m_2}{m_1} \text{Log}_n X_1 - \text{Log}_n X_2}{\frac{m_2}{m_1} - 1} \quad (13)$$

m_2/m_1 is proportional to the thickness for each area, thus equal to the ratio of the number of layers of metal in each measured pixel. Table I gives the measured value for K in different areas and this together gives the constant needed in equation (12) and validates the method since this value is constant with a good accuracy. This constant is a characteristics of the operation conditions and is determined once only. It is necessary to re calibrate only if the exposure or development conditions are changed. Table II presents the results in a different manner. From table I a mean value for K has been calculated as -1.53, then equation (12) is computed using this constant for different areas. This gives an idea of the accuracy of the method. It must be emphasised that the computer computes automatically the equation and presents the result as an image where the mass changes are directly visible. There is no need to compare the pixels values to a standard for each pair of radiograph as in DSR.

All explanations have been given using equation (12). The method is the same if the optical densities are high enough so that one must use equation (10). The constant C is determined as the constant K .

DISCUSSION

Theoretical limitations

We assume that the coefficients of equation (2) are the same for all wavelengths which is true (18).

The camera is assumed not to introduce any geometrical deformation, and the light used for the digitisation adjusted so that the dynamics of the camera is in the range where it is not saturated. It means that the illumination must be uniform in all the field of view and adjusted in a “reasonable range” nor too low nor too high. We have checked the characteristics of the camera as explained. If the response was not linear it would be possible to put the measured curve in the memory of the computer to correct this problem.

The mass ratio equation (10) or (12) are valid for a single wavelength λ only. Far from an absorption edge, μ/ρ varies as λ^3 . The X-Ray beam is polychromatic but all the calculations have been made for a monochromatic wave only since no analytical calculation would be possible otherwise. However since most of the energy is contained in the characteristic X-Ray emission wavelengths of the X-Ray tube this is a valid simplification which has been verified when validating the method. This simplification is acceptable for the wavelengths used in medical radiography (16).

By comparing areas without changes in various radiographs i.e. areas where the mass ratio should remain 1, we have been able to estimate that the best precision is of the order of 5%.

The difference technique:

Many authors (1-9) have computed the difference between two digitised images (DSR) to analyse changes in bone mass. After adjusting the contrasts between two radiographs, they compute the difference between the pixel values p_2 and p_1 . This difference is a function of μ/ρ and does not consider separately the cases where the optical densities are either in the linear or non linear part of the densitometric curve.

Quantitative information requires calibration with a standard. Moreover the standard is not the same for different tissues since the mass absorption coefficient μ is not the same. This is not true for DIR since μ is not present in equation (10) and (12).

We suggest that Fourier filtering could be used for DSR as we have done for DIR to eliminate the noise which is always additive.

Influence of illumination geometry

Like other methods, digital image ratio (DIR) is sensitive to variation in the illumination geometry because each pixel of the first radiograph is mapped into another pixel in the second radiograph. Thus the same conclusions are valid for DIR. We assume that a tilt of $\pm 2^\circ$ or $\pm 3^\circ$ is acceptable (19, 20).

Operating parameters

X-ray machines are usually operated between 70 kV - 90 kV to obtain the best visual image contrast. Because the radiographs of this study are intended for digital methods, they do not need to be visually pleasing. The X-ray machine was operated at 50 kV. X-Rays absorbed by bone correspond to an energy range between 11.5 keV and 23 keV (fig. 3). When operating the X-Ray machine at 50kV the maximum intensity X corresponds to an energy of 22.8 keV which is the upper limit of the absorbed energy. Increasing the voltage mainly shifts the emitted energy spectrum to higher energy. Thus it is useless to operate the tube at a higher voltage. Increasing the voltage also increases the diffuse scattering which appears as background in the image and is a source of error for a quantitative measurement (21, 22). The sensitivity of the X-ray films decreases when the energy of the photons increases since the most energetic photons pass through. According to Okano (23) a Kodak D film exposed at 70 kV has a scattering section of 0.1, i.e. 10% of the photons only are absorbed. It is not efficient to use a too energetic radiation.

CONCLUSION

DIR does not suffer from certain difficulties encountered in other methods (8-11) for comparing superimposable radiographs. Diffuse scattering arising from soft tissues and other dental structures need not be taken into account, as it is the same for a given patient and site. The spectral emission characteristics of the X-ray tube are not important, as long as the same machine operating in the same conditions is used. The numerical noise, which is a major source of errors, is removed by multiple acquisition and Fourier filtering.

DIR has the advantages of showing the density of matter by displaying directly the mass ratio. DIR is a simple method which can be used for long-term studies. It combines the advantages of low X-ray machine voltage (50 kV) and short exposure times, making it possible to study several sites on the same patient with lower irradiation doses.

DIR has been successful to measure bone change. A full clinical study is presented in part II which validates this new method.

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Appendix A

Contrast adjustment

Let $p_1(i)$ be the density of a pixel in the first image, $p_2(i)$ the density of the pixel in the second image at the same location i . The image contains N pixels, numbered from $i = 1$ to $i = N$. In the present case $N = 512 * 512$.

We compute:

$$\bar{p}_1 = \frac{1}{N} \sum_{i=1}^N p_1(i) \quad \text{and} \quad \bar{p}_2 = \frac{1}{N} \sum_{i=1}^N p_2(i)$$

$\bar{p}_1$ and $\bar{p}_2$ are the mean densities of the images. If changes occur in limited areas only, computing the densities in the whole image does not introduce significant errors. Otherwise these statistics must be computed from areas where one is sure that no change occurs.

Then we compute the variances :

$$\sigma_1 = \frac{1}{N} \sum_{i=1}^N (p_1(i) - \bar{p}_1)^2 \quad \text{and} \quad \sigma_2 = \frac{1}{N} \sum_{i=1}^N (p_2(i) - \bar{p}_2)^2$$

The relation between the densities of the pixels is linear, since it corresponds to the linear part of the densitometric curve:

$$p_2(i) = a.p_1(i) + b$$

thus:

$$a = \sqrt{\frac{\sigma_2}{\sigma_1}} \quad \text{and} \quad b = \bar{p}_2 - \sqrt{\frac{\sigma_2}{\sigma_1}} \bar{p}_1.$$

This is a statistical method where all the pixel values are modified through a linear equation. The information in the image is not modified. Other methods such as histogram equalisation modify locally the pixel values and the information is irreversibly changed. The image may look nicer but becomes quantitatively wrong.

m_2/m_1	2/3	7/10	9/7	4/3
K	-1.45	-1.55	-1.57	-1.55

Table 1. Values of the constant K (equation (12)) as calculated from different areas of the step wedge. First line: value of the mass ratio, second line: constant as calculated from equation (13).

m_2/m_1 theo.	2/3	7/10	9/7	4/3
m_2/m_1 calc.	0.65	0.69	1.28	1.32

Table 2. True mass ratio and corresponding calculated mass ratio for different areas of the step wedge, using the mean value from table 1 for K.

Fig. 1. Flowchart of the digital image ratio method (DIR).

Fig. 2. Typical response of a photographic emulsion.
 (a) Density of grey versus the intensity of the incident beam.
 (b) Same curve plotted on a logarithmic scale.

Fig. 3 (A) Change in the mass absorption coefficient of calcium as a function of the incident X-Ray beam energy.
 (B) Same curve for hydroxyapatite
 (C) Spectral distribution of X-Ray radiation for a machine operating at 50 kV.