Spatial Resolution of Cephalometric Radiographs

by

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Abstract

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The present study examined the spatial resolution requirements necessary to use digital representations of lateral cephalograms in the same way as clinicians and researchers now use lateral cephalometric radiographs on silver halide cellulose This study used twenty five lateral cephalometric radiographs chosen at random from the 428 radiographs used to develop the Bolton Standards. These 25 films were photographed and transferred to a photographic compact disc Five images were chosen randomly from the sample of 25 and used in a subjective test on spatial resolution. Using two matched monitors, a spatial resolution test was devised to compare the same radiograph at three different levels of spatial resolution. The results from twenty eight operators indicated that an optimum spatial resolution of lateral cephalometric radiographs was 768 pixels X 512 pixels

The twenty five original radiographs were traced twice by four examiners and once by a fifth examiner. An objective test of the spatial resolution was then carried out by five judges locating landmarks on the original radiographs, and comparing this data to the landmarks located off the digital image, shown at a spatial resolution of

768 pixels X 512 pixels All landmarks were located in a x,y coordinate fashion, using a double determination no sooner than two weeks apart.

From the original twenty five lateral cephalometric radiographs and from the twenty five electronic images twenty one anatomic and four fiducial landmarks were located. The landmark location was compared between the different media, different examiners, and at different times Landmark location overall from the computer monitor was found to as accurate as that from the conventional cephalometric silver halide radiographic technique. Individually, some landmarks were more accurately located on the computer monitor, while others were more accurately located from the conventional radiographic technique. An exception were the fiducial landmarks which were significantly more accurately located from the conventional cephalometric technique. Recommendations were made to use the 1536 pixel X 1024 pixel images and to investigate gray scale resolution of cephalometric radiographs.

To Denise, Macree, John Francis, and Siobhan
Thankyou for your faith, inspiration and guidance

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INTRODUCTION

Orthodontists have traditionally used a standardized database of information to diagnose and plan their patient's treatment. The existence of comparable data sets has also allowed orthodontists to determine the response to various treatment interventions in a scientific manner. Plaster casts of the teeth and cephalometric radiographs have been the mainstay of orthodontic records. Fortunately, before radiographic equipment gained wide acceptance in clinical practice, Broadbent (1931) established a standardized, repeatable method for obtaining headfilms. Since then, orthodontists have been fortunate that all radiographs taken on patients share some common features. Standardization was important because it allowed all orthodontists to work with similar information. The cephalometric method has allowed orthodontists to progress at a much faster rate because data could be shared among colleagues.

At the recent American Association of Orthodontists national meeting in Toronto (May 1993), there was a virtual explosion of new computer products capable of storing and /or analyzing orthodontic records. With the pressure that is being exerted by OSHA to eliminate cross-contamination in the orthodontic workplace and the relative ease with which computers can now handle image based information, it is easy to see that the paperless office is not far away Unfortunately, the introduction of

computer technology into the orthodontic office has occurred before the analytic requirements of a computerized cephalometric radiograph have been determined.

Currently the profession is being challenged to provide more information on the efficacy of various treatment methods This challenge has increased the need for a standardized database of treated and untreated orthodontic cases. Because of the high degree of variability in human malocclusion, a very large database is needed. A sufficiently large database would allow selection of groups of cases that are matched on enough variables to allow scientific validation of various treatment interventions Historically, the rate limiting step in establishing large databases of orthodontic information has been the time required to copy records and store them at a common site. The personal computer may provide a solution to this problem because it is relatively easy to make copies of computerized information. However, for the copies to be useful they must be compatible among various computers and computer software programs.

To continue to communicate effectively in the information age, we need to establish a standard way of storing and retrieving computerized cephalometric radiographs. To date no standards have been proposed or adopted by the profession for storage and exchange of computerized patient information. One of the issues that need to be addressed before standards can be established is that of resolution. What is the resolution (spatial and gray scale) necessary for orthodontists to use digital representations of lateral cephalometric x-rays instead of the original films?

The purpose of this line of research is to determine the spatial resolution requirements necessary to use digital representations of lateral cephalograms in the

same way as clinicians and researchers now use lateral cephalometric radiographs on silver halide cellulose.

The specific aims of this study are to determine the spatial resolution necessary to.

Subjectively evaluate various operators ability to discriminate between the digital representations of the same radiograph at three spatial resolutions shown in a A B comparison on the computer screen;

Objectively identify twenty one common cephalometric landmarks and derive reproducibility indices for the digital media compared to the gold standard original silver halide radiographic film.

Our goal is to define the needs of the profession now, before the majority of orthodontists are using computerized record systems. By standardization of record keeping and display of patient information, compatibility of records can be ensured. Consequently, this will allow comprehensive databases of treated and untreated cases to be established in the near future. Such databases will provide a network of useful information about clinical orthodontics. With the long history of standardized record taking in orthodontics, it would be ironic if the technology that will allow assembling large databases of information lacks the standardization necessary to utilize these databases optimally. In short, orthodontists need to be able to exchange computer disks that contain stored records with the same ease they now exchange cephalograms and models.

Literature Review

The Resolution Problem

Before a method is chosen for electronic capture of cephalometric radiographic images, a number of decisions need to be made. The principal decisions are the method of image capture, and the resolution of the image. It is imperative that whatever methods are chosen, the electronically captured image must have the same information stored in it as the original radiographic film. The two major parameters which affect the quality of the scanned images are:

1. The spatial resolution; and
2. The grayscale resolution (Hans, Thomas and Broadbent 1992).

Resolution is a difficult concept to comprehend in its true definition. Resolution is defined as the act or process of resolving or separating into constituent or elementary parts (Random House College dictionary 1984). The true definition of resolution does not assist one in understanding the concept of resolution in digital image processing. However, the following explanation from Dr. Mark G Hans, Dr. Cecil Thomas, and Dr. B H. Broadbent (Unpublished paper 1992), enables one to understand the meaning of resolution in terms of digital image processing.

> Traditionally, resolution refers to the quality of an image in terms of the fidelity of the finest details in the content of the image. Thus, the spatial resolution refers to the image quality in terms of the spatial detail. When the spatial resolution is poor, two adjacent spatial features may be blurred together so that they appear as a single feature rather than two separate features. This classical definition of resolution is obviously dependent on the image content, and on the importance of small features as carriers of image information. To make resolution measurement of an image more specific, spatial resolution is often specified in terms of "lines per inch" or "line pairs per inch". This nomenclature refers to the maximum number of

alternating black and white lines which can be seen in an image under repeatable calibration conditions. The term "lines of resolution" is also used. In digital images, an image-independent and display-independent specification is simply the number of pixels in the horizontal and vertical directions. While this is not the only determinant of spatial resolution (in terms of resolving the details in an image), it has become a standard measure.

Grayscale resolution refers to the fidelity of an image (or imaging process) in the accurate representation of the brightness and contrast scale in an image. In an analog image, this scale has a large number of gray levels. In the noiseless case, the number of gray levels is infinite. In a digital image, the number of gray levels is an integer, usually two to some integer power. When there are few gray levels utilized in an image, a gradual change in image brightness becomes a stepwise change. In two dimensions, the steps become contours which separate regions of constant brightness.

In terms of cephalometric radiograph scans, what is the significance of the spatial and grayscale resolutions? Suppose we wish to locate a single landmark, as is necessary to make certain distance measurements. If the spatial resolution were too low (i.e., insufficient number of lines per inch in an analog image or horizontal and vertical pixels in a digital image), then the image would appear to be blurred. The blurring may completely obscure a landmark, and the measurement would be impossible. Less severe blurring would make the landmark less "sharp", larger, and merged with neighboring features. This blurring introduces ambiguity and error in the measurement and the location of the landmark. Given the original film images, some minimum number of lines per inch (or horizontal and vertical pixels) are required in a scanned image, so that the information content of the film is transferred to the video or digital image. The information content can be obscured or lost by insufficient grayscale resolution. A digital image scanned by a video camera may have 7 bits grayscale or 128 discrete gray levels. Better quality (and more expensive) cameras may have 256 grayscale levels. Many texts and sales persons state that 64 gray levels are sufficient for human viewing. However, a simple experiment can show that we easily see a difference of one in 256 gray levels. Thus, the requirement is highly dependent on the image content and the actual measurement made on the image. In terms of the cephalometric radiograph scan, if a landmark feature is in an image region with relatively slow and perhaps subtle changes in grayscale, then too few gray levels may obscure the landmark or change its apparent location

Unfortunately, there is no theory or equation to calculate exactly how many pixels and gray levels to use. Generally, the maximum number of gray levels available are utilized. Furthermore, the grayscale level and number of pixels are not independent. For example, the number of horizontal and vertical pixels may be adequate for 256 gray levels, but at 64 gray levels, a superior spatial resolution may compensate for the loss of grayscale resolution. This trade-off between spatial and gray scale resolution is limited. If an image has too few gray levels, increasing the spatial resolution cannot compensate for the loss of information content.

Once a minimum spatial resolution has been determined, the stage will be set for future research to examine the question of optimum gray scale resolution. The spatial resolution recommendations will be independent of a particular company's

computer hardware or software That is to say, the standard will be for the image resolution, not the computer or monitor that is used to display the information. Furthermore, the software used to measure the radiographs can also vary between orthodontists The important point here, is that a practitioner can store the cephalogram on a computer disk in a standard format that can be given to another practitioner that may use a different computer and a different software measurement program. However, because of the standard, the image on the disk will be able to be used just like a practitioner might now send a cephalogram to a colleague to gather input on treatment planning.

Computers and Radiology

At the first Picture Archiving Communication System (PACS) meeting in 1982, the medical profession was aware of the problem of acquiring data from different manufacturers equipment was already apparent Each independent manufacturer of imaging equipment was using their own proprietary software and file formats to acquire and store digital medical images. The vendors were not keen to disclose any details to enable sharing of image files from one format to another. Consequently, the medical profession consulted the users of PACS, and in conjunction with the American College of Radiology, formed the American College of Radiology - National Electronic Manufacturers Association (ACR-NEMA) Digital Imaging and Communication (DICOM) Standards Committee in early 1983 (Horrii et. al. 1990).

The original objective of the Committee was to develop an interchange system enabling transfer of medical images from one computer to another computer of a different type (ACR-NEMA 1984, from Horrii et. al. 1990). The initial task was to examine different file transfer protocols currently available Those examined included: Institute of Electrical and Electronic Engineering (IEEE) 488, Ethernet, Intelligent

Peripheral Interface, Small Computer System Interface (SCSI), Digital Equipment Corporation (DEC) DR11W, DEC Unibus, and Multibus. Standards for data formats included American Association of Physicists in Medicine (AAPM) Magnetic Tape, LANDSAT, NATO magnetic tape, and the Japan Association of Medical Imaging Technology (JAMIT). Although some ideas were melded from all of these, not any particular one was chosen. In 1985, the NEMA Standard (ACR-NEMA 300-1985) was published (Correia et. al. 1990).

The ACR-NEMA modified the American National Standards Institute (ANSI) standards for magnetic tape. Using the ANSI standards it was possible to fit the ACR-NEMA data set structure in the ANSI standard file structure with a wrapper consisting of ANSI labels. Later in 1985, the committee was considering a standard for erasable optical (magneto-optical) disk (Correia 1990).

Historically, nuclear medical images were the first to be regularly digitized because they have relatively low resolution and contrast compared to other medical images (Hounsfield 1973) One must remember that 15 years ago, computer hardware available was quite limited in both speed and capacity, such that only low information content images could be efficiently handled in the digital regime. X-ray computed tomography (CT) of the early 1970's was a digital modality in the sense that an output was a computed image. CT scanning was the first modality to require a computer for image production and it naturally followed that the digital images produced would be displayed electronically This set the stage for the digital imaging revolution taking place today For the first time radiologists were exposed to the advantages of post acquisition image manipulation and began to accept how useful this process can be.

Since the introduction of CT, there have been several developments in digital imaging. The first includes computer dependent modalities such as magnetic

resonance imaging (MRI) (Lauterbur 1973, and Hinshaw et. al. 1978). The second type involves the application of digital image acquisition and display methods to intrinsically analog modalities such as planar radiography and fluoroscopy (Sackett et al. 1980). This second group of applications is also called digital radiography. The third development was real time ultrasound (Wells 1977, and Wells 1980).

Computers and Cephalometrics

Computers in cephalometric research have an interesting and varied history. One of the first applications of computers in the field of cephalometric analysis was to reduce the amount of manual work required in the statistical calculations. In the late 1950's lateral cephalometric radiographs were translated to a coordinate model, which was stored on computer punch cards (Walker 1972). As early as 1964, lateral cephalometric diagrams and printouts were being computer generated (Walker 1972). Computer assistance soon led to extensive analyses of the properties of correlation between cephalometric variables (Bjork and Solow 1962, and Solow 1966). Later, factor analyses were introduced These extensive factor analyses required considerable computing to manage the numerical data manipulation (Brown et. al. 1965, and Solow 1966).

In the early 1970's many computerized applications were developed to extract data from cephalometric films (Ricketts et. al. 1972, Solow 1970, Walker 1972, and Chebid et al. 1976). An early digitizing system for analyzing cephalometric radiographs was described by Bergin et. al. (1978). In their study they found the greatest error was not in the digitizing process, but in the location of the anatomical landmarks. Accuracy was excellent in transferring data from the tracing film to the computer Baumrind and Miller in 1980 described the UCSF computerized cephalometric method. Their method was well thought out, and had been

implemented successfully. The goal of this technique was to evaluate the effects of orthodontic and orthognathic surgical treatment through time, and focused on measurement reliability, intra-individual comparisons at different time points and the evaluation of grouped data This was and is in contrast to many of the published cephalometric techniques that were mainly dealing with prediction of treatment planning or growth forecasts These techniques usually disregarded cephalometric error, and proclaimed growth prediction from single measurement of an individual cephalometric films.

Baumrind and Miller (1980), reflecting on ten years of experience in computer management of cephalograms, offered many suggestions for future orthodontic computer packages. Recognizing the principal problem of utilizing cephalograms is that of landmark identification, many issues were discussed on reducing error and increasing reliability in measurement and landmark location It is unfortunate that commercially available cephalometric packages did not heed the wisdom of Baumrind thirteen years ago To date, not a single company selling cephalometric software has implemented the basic recommendations of Baumrind One of Baumrind's outstanding concepts was placing fiducial reference marks on lateral cephalometric films. Fiducial reference points are commonly used in the engineering discipline of photogrammetry. All the landmarks located, and the measurements made could be related to these fiducials as Cartesian coordinates. Hence, films can be accurately superimposed on each other, and the amount of error (which is highly significant), can also be determined. Baumrind (1972 and 1980) also presents the logic for doing multiple tracings, or digitizing individual films The probability of examining one film (20 landmarks) without making at least one error at the 0 05% level is 36% Whatever the error is in estimating an individual landmark, it may be halved by using the mean of

four independent tracings. For clinical treatment planning decisions, at least two tracings are recommended, and for research four tracings (Baumrind 1980).

Broch et al (1981), considered the reliability of measurement to be a cause of major concern particularly in studies of single individuals, or where the measured dimension was quantitatively small. They suggested the use of high quality equipment and a precise method of recording as a means to decrease the total error associated with the process. Richardson 1981 notes the digitizing method holds only slight advantages over manual tracing, and the difficulties lie in pinpointing poorly defined landmarks in X-ray plates, where their location is often subjective

Bondevik, Rosler and Slagsvold (1981), outlined a rudimentary digitizing computer appliance A center point of a ruler was placed over a landmark, and this point was digitized, then the ruler was moved to a new landmark and another point was digitized. A manual system was used to measure the angular relationship between various cephalometric landmarks. Other investigators have examined the possibility of computer recognition and identification of skeletal landmarks. This line of thought has been used to attempt to overcome the main source of error in cephalometrics which is landmark location. One group carried out their testing on a Clip 4 computer This machine had a very low scanning resolution (compared to our standards today), of 96 X 96 pixels, and a gray scale resolution of 64 shades of gray, alternatively known as 6 bit images They speculated that the low resolution obtained, limits the reproducibility of results (Cohen, Ip, and Linney 1984). An interesting use of using three dimensional x-ray stereometry from paired coplanar images utilizing modern mapping techniques has been presented (Baumrind, Moffitt, and Curry 1983). Other investigators have extended the use of conventional algorithms of photogrammetry by use of vector

geometry to assist in the location of reference points from a pair of biplanar cephalograms (Brown and Abbot 1985, 1989).

Cohen (1984), found tracings made prior to the digitization process reduced the reliability of measurement, and that uncertainty in measurements lay with the correct location of the required landmark Others reported on their use of image enhancement techniques as aids to the correct location of cephalometric landmarks. Using three enhancement modes, a Sobel gradient in combination with median filter, a statistical difference filter, and a tracing enhancement (Sobel output plus original), they found no statistical improvement in using image processed methods (Jackson PH, Dickson GC, Birnie DJ. 1985).

Future Recommendations

Cephalometric landmark location is typically performed by trained human judges on standardized lateral skull X-ray images or on photographically duplicated copies of such images. Such images may be considered to be of the continuous analog type, both with respect to spatial dimensions and with respect to gray scale. Digital imaging will become the primary method of cephalometric appraisal in the next decade as:

1 It is easier to provide several identical digital copies of same X-ray image at different locations than it is to provide identical photographic copies;

2. Digital images are much more amenable to automated image manipulation, enhancement, and analysis, than are photographic copies,

3 It is highly likely in future years that X-ray images will be captured in a digital form directly, completely bypassing photographic film This should allow reduction of the radiation dose, and increase the gray scale dynamic range Note that the electronic sensors currently in use with Computed Tomography (CT), have a gray

scale dynamic range of $10^{6,}$ as opposed to the 10^{3} physical limit of modern X-ray film screen combinations (Baumrind S, B Baker, W Hane and G Nazareno Waiting on Pub). Others believe the gray scale dynamic range of modern radiographs to be in the region of 10^{4} (Hans et. al. 1992)

4 Digital images may be archived and retrieved efficiently from digital data bases (Correia et. al. 1989).

Computer Graphic Files

There are many different graphic file formats, but only two kinds of graphics· *bitmapped graphics*, also known as *pixel-based graphics*, and *vector-based graphics*, also known as *object-oriented graphics*. The differences between these two types of graphics involve the kinds of software applications that produce them, the way the computer stores the graphic in memory, and the ease with which you can edit the graphic.

Scanners and paint programs produce bitmapped images by establishing a grid of pixels, the smallest visual unit that the computer can address on the screen Scanners and bitmapped software applications create images by altering the colors or attributes of each individual pixel. This way of storing information makes inefficient use of memory The size of an image is the number of pixels it occupies, and this is fixed once an image is created. The number of pixels created is dependent on the resolution of the computer display where the image was first created. As a result, finished bitmapped images are difficult to edit when you transfer them from one application to another. If the size of a bitmapped image is increased, white spaces and jagged edges may appear. If the size of a bitmapped image is greatly decreased, parts of the image may smudge due to the compression involved. Distortion of the image

can also occur if a bitmapped image is transferred to another computer that has a different display resolution.

Vector based images, or object-oriented graphics, are stored in the computer's memory as a series of numbers, not pixels. This series of numbers describes how to redraw the image on the screen The method of storing Object oriented graphics, has nothing to do with the resolution of any given display adapter, and is considered to be device independent No matter what computer is used to create an Object oriented graphic, the graphic can be stretched, scaled and resized without distortion. Object oriented graphics also tend to create smaller files than bitmapped graphics because the computer does not have to memorize the attributes of individual pixels.

Common Bitmapped Files

TIF file extensions denote Tagged Image Format files. These are bitmapped graphics and are generated by scanners and some software. PCX and PCC file extensions are native to Z Soft PC Paintbrush, and these kinds of bitmap files are produced by some scanners and popular paint programs. Microsoft Windows bitmap files have the BMP file extensions The BMP files are generated by Windows 3.x applications, and also by the screen capture utility in Windows. Windows BMP files will not convey gray scale information.

Specifying Bitmapped Graphics resolution

Computers store pixel based graphics as individual units. It is most important to include only the information necessary, as otherwise extensive memory will be used up. When utilizing a pixel based graphic, use the minimum size of the image that will be required for the final application.

Digital Image

Analog representations of images have been used long before numerical ones. Currently, storing, treating, and transmission of information is done more and more in numerical form, both in the case of sound (digital telephone) and in the case of images (digital television). The reason for this is that digital information can be processed rigorously (logical and arithmetical operations on a computer), and they can be stored and transmitted without loss or distortion. This is not the case for analog information. An image can be considered as a function I=I[x,y] where x and y are the coordinates of a point of the image with respect to a Cartesian coordinate system and where I is a function giving light intensity and color: I=[R(x,y), G(x,y), B(x,y)] Digitizing the image means to sample the signal I on a grid of step sizes Dx and Dy In every surface element of sides Dx and Dy, an average value of the light intensity there is realized, which is digitized as a word in the screen memory. The decoding is done by displaying at the point of coordinates (x,y) of the screen pixel of dimensions Dx and Dy of the intensity of the stored value (Brett 1991).

A pixel or picture element is the basic display unit of a plane film In the case of a cross sectional, volumetric modality such as CT it is called a voxel (volume element).

Scanning

Scanning involves viewing an image and converting the image into a digital format. Various scanning resolutions are possible, the actual value will vary depending on the individual scanner used. With most scanners commercially available, one can adjust scanning resolution, brightness and contrast. One can also prescan the image to choose a particular area for scanning, this is known as cropping an image. This is quite a useful feature, as only the area required is scanned, and this limits the

time taken for the scan and also the amount of disk space required for the image. Whatever way an image is scanned or created, it is transformed into a collection of color and or gray values. These color/gray values are stored in one of several image data types.

Based on the display configuration of the hardware utilized, and the types of image formats being used, one may need to convert the image from one format to another. Most image capturing devices have difficulty differentiating among fine differences in image brightness - they do not sample the whole range of contrasts or support enough dynamic range to properly record all the brightness differences. This can make it difficult to bring out the highlights, shadow areas, or midtones of an image. If you compensate by changing the scanner contrast and brightness settings, scanners often produce noise (extra pixels) in the information they capture.

Image Types

Data types differ in the number of bits per pixel that are used to record information about the pixel and the number of channels [color components] that comprise the image. The number of bits per pixel determines how many colors or shades of gray can be represented by each image pixel. The ability to manipulate an image depends upon its data type.

Black and White

Each bit of a black and white image is governed by one bit of information. This one bit can have a value of 0 [black] or 1 [white].

Gray Scale

Each pixel of a gray scale image is controlled by eight bits (normally), whose value determines the gray intensity, i e 256 shades of gray. This works in a logarithmic fashion

1	bit image		black and white
2	bit images	4	possible gray shades;
4	bit images	16	possible gray shades;
5	bit images	32	possible gray shades;
6	bit images	64	possible gray shades;
7	bit images	128	possible gray shades;
8	bit images	256	possible gray shades,
9	bit images	512	possible gray shades;
10	bit images	1,024	possible gray shades;
11	bit images	2,048	possible gray shades;
12	bit images	4,096	possible gray shades;
13	bit images	8,192	possible gray shades;
14	bit images	16,384	possible gray shades;
15	bit images	32,768	possible gray shades,
16	bit images	65,536	possible gray shades,

256 shades of gray is comprised of values ranging from black to white, where 0 represents black, while 255 represents white.

Image Conversion

To produce gray shades, computer software divides the original image into square cells, the size of these cells varies from 1 pixel by 1 pixel to 16 pixels by 16 pixels. Each cell (for 8 bit) can contain any of 256 shades of gray (this includes black and white). The software then determines from the original image what shade each cell should be by taking an average form the original black and white pixels in the cell area, it can also interpolate between areas of black and white pixels to introduce new pixels with different gray shades.

Digital Radiographs

A digital image of a radiograph is produced by creating a rectangular array of numbers representing gray values (or shades of color for color images) from the negative film image. The elements of the array are termed pixels, in a 12 bit image each pixel may be represented by 4096 shades of gray. As a comparison, a television set can display only approximately 25 gray shades By convention, the pixels are

integers in the range of 0 to 4096, with 0 representing black and 4096 representing white. The image display part of the computer system converts the numbers into a video picture. Image processing is related to the visual effects of performing mathematical operations on these numbers (Eppley and Sadove 1991).

The radiological information has been conventionally recorded on film, because until recently only film could record the fine detail, wide dynamic range and large area in a sufficiently short time. But film is becoming more expensive to use, bulky to store, and inconvenient to distribute. The information content in a radiological film is difficult to manipulate. This contrasts with other medical imaging techniques of computed tomography, digital subtraction angiography and radioisotope imaging which are being recorded in digital form.

Dental researchers have used digital processing systems to examine a variety of questions relating to caries detection, the evaluation of periodontal bone height, detection of various jaw conditions, performance of tracing on cephalometric radiographs and forensic teleradiology (Jackson et. al. 1985, Okano et. al. 1982, Grondahl et. al. 1982, Pitts 1984, Grondahl et. al. 1984, Van Der Stelt et. al. 1985, and Southard and Pierce 1986). Medical researchers have investigated the effect of digitizing format, specifically pixel size, on the detectability of various lung conditions (Cocklin et. al. 1983, Mac Mahon et al. 1986, Fujita et. al. 1988, Foley et al. 1981, Fraser et. al 1983, and Goodman et al 1986), and more recently gastrointestinal conditions (Kastan et al. 1987).

One of the many advantages of storing the X-ray images in digital form is that it is then possible to process the image data using the full range of image processing techniques now widely used in a variety of other application areas. These include such

diverse subjects as satellite image processing, geological surveys, and surveillance (Cocklin, Gourlay, Kaye, Miessler, Kerr and Lams 1984)

Some investigators believe there is more information available in a digitized radiograph than on a conventional radiograph because computer manipulation of the image permits adjustment of contrast, density, edge enhancement, and also permits enlargement without loss of image quality (Eppley and Sadove 1991). This last statement certainly could only be correct if the captured image had enough spatial and gray scale information stored in it. Once an image is digitized with a given pixel size its unrecorded detail is lost forever. The optimal spatial resolution for a given application provides for an acceptable level of diagnostic accuracy while reducing data storage requirements to a practical minimum (Kassebaum et. al. 1989).

A digital image permits a wide range of processing methods. One of the more useful techniques involves modifications of image contrast and brightness These modifications can be used selectively in a small region of the radiograph so that detail can be more easily perceived e.g. one may have higher contrast in the lighter areas of the image, and less contrast in the darker areas In addition various structures can have their edges enhanced, thereby improving their visibility. Edge enhancement, like contrast and brightness, can also be selectively applied to either darker or lighter areas or to the entire image. The amount of applied edge enhancement can also be varied The net result of computerized digital imaging is that one can consistently acquire images of uniform density with selective enhancement of any particular structure within the image. In particular this means that soft tissue will be more easily visualized on digital films than conventional films. Being able to predict accurate soft tissue change is essential for growth monitoring and documenting post orthodontic treatment change (Eppley and Sadove 1991) However, some investigators experimenting with

fast film screen combinations, concluded that although fast systems were qualitatively inferior to slower systems, there was little difference to the actual measurements taken on cephalometric analysis (Halse and Hedin 1978). This may indicate that film quality is not directly proportional to cephalometric analysis with experienced operators.

Resolution Requirements

The minimum number of elements needed to represent a digital image from a given modality depends on the intrinsic spatial resolution of that modality. This relationship arises from the need to have sufficient elements to represent the image without degrading its intrinsic resolution and is dictated by a fundamental result of signal processing theory called the sampling theorem (Barret and Swindell 1981) The sampling theorem requires that for a given spatial resolution, R, defined by the physical limitations of the modality, it is necessary to have at least two image elements per resolution length to preserve the information in a digitized image In practice, more than this number of elements is needed For example, if a given image has a characteristic dimension I, and a resolution of R, then a matrix with a dimension of at least 2I/R is required (Correia et al 1989). Correia also gives an example of a 30 cm digital radiograph requiring a 1500 X 1500 pixel matrix (minimum 2048 square). Hence for our cephalometric films, the 3000 X 2000 scans seem quite appropriate

If a digital image does not meet the minimum spatial sampling requirement imposed by sampling theorem, then information is lost in the digitizing process. This may result in a phenomenon called aliasing in which artifacts or distortions are introduced into an image. In practice, it is sensible to choose a matrix size that is sufficient to represent the given image and not a larger one. The larger a matrix, the more mass storage it requires and the longer it takes to store, retrieve, transmit, and manipulate the image (Correia 1989).

One of the factors that has limited the digitization of high resolution planar radiographic images is the storage required to digitize these images at full resolution is very large. For example, to preserve all the resolution in a chest film having 0.2 mm intrinsic resolution requires a matrix image size of 4096 X 4096 (Correia 1989). However, a matrix size large enough to maintain the full resolution is not always necessary as the information content of the measured data rather than the resolution may be the limiting factor. Factors such as image contrast and structure size may contribute to defining information content (Correia 1989).

The contrast resolution of an image is qualitatively defined as the fineness with which differences in the imaged parameter can be detected for a given image at a given resolution. An image may be described by the number of elements and the number of bits of contrast needed, for example, 512 X 512 X 12 bit. As in the case of spatial resolution, under sampling of the contrast range of an image can cause artifacts. The most typical artifact is the appearance of false edges or contours in an image. This can be confused with true edges and structures resulting in errors in clinical findings (Correia 1989) A radiographic film image requires a contrast resolution of 0.2%, this means at least 1000 gray levels are required (i.e. 10 bit) (Correia 1989)

This issue has not as yet been adequately tested in the handling of lateral cephalometric radiographs It seems that in the dental literature the maximum resolution possible at that time was used for any system described However, digital radiological imaging has been used in medicine for a number of years. The image resolution requirements are dependent upon the level of anatomic or pathologic detail to be retained. A vertical resolution of 280 lines may be adequate for demonstrating fractures, while for diagnosing slight density variations in the mediastinum the acceptable resolution may exceed 1000 vertical lines (Tesic et. al 1983) In describing

a digital image processing system designed in 1984, for chest X-ray, the authors constructed a system that could capture image sizes of 1024 X 1024 pixels at 16 bits per pixel (Cocklin et. al. 1984). These images were of a high resolution for that time period

In 1984, it was felt image sizes greater than 1024 X 1024 pixels for chest radiology, were not needed (Foley et. al. 1981 and Cocklin and Lams 1983). But, it was stated for digital imaging to be effective, the system must be able to accommodate the entire area normally captured with X-ray film, and achieve the diagnostic accuracy of film screen combinations (Cocklin et. al. 1984) However other investigators felt it was not until a matrix size of 2048 pixels X 2048 pixels was achieved, that a digital radiograph would have the equivalent spatial resolution of that of an analog image stored on X-ray film (Huang 1987). However, Jackson in digitizing cephalometric images using resolutions of 512 pixels X 426 pixels and 1024 pixels X 852 pixels (both at 8 bit gray scale per pixel) found the 512 pixels X 426 pixels images were only marginally inferior to the higher resolution of 1024 pixels X 852 pixels. Yet, in his conclusion Jackson recommended increasing the spatial resolution to 1024 X 852, and increasing the grayscale level to 9 or 10 bits per pixel (Jackson et. al. 1985). Approximately 1500 X 1500 pixels are required for chest radiology (Cocklin et. al. 1983) and this is equivalent to 600 to 800 square pixels for cephalometric films due to their smaller size (Jackson et. al. 1985) A lateral cephalometric radiographic film is approximately 10" X 8" or 20 cm X 25 cm.

In diagnostic radiology, digital images often requires a matrix larger than 512 pixels X 512 pixels for processing and display However, one may combine four 512 images to form one 1024 image. Alternatively each 1024 image can be stored as four 512 images (Huang 1987).

In order to display a 1024 image on a 1024 monitor, the image master synchronization clock must have its fundamental rate increased. A pixel interleaving scheme can be devised which spreads adjacent pixels over four memory boards and adjusts pixel addresses accordingly (Huang 1987) Currently, the most common method of displaying high resolution images is with a non interlaced display, using high refresh rates, and high memory 24 bit video cards (perhaps combined with a graphics coprocessor).

Capturing Digital Radiographic Images

Video Scanning

Early technology used a television camera to obtain an analog image directly from the radiograph. The analog image acquired was subsequently digitized. This system had limited usefulness as image degradation was severe. The limiting factor was the resolution capability of the television camera. Early dental research in video capture techniques followed the digital subtraction radiography technique of Buonocore et. al. 1981 This method allowed immediate viewing of dynamic changes, however spatial resolution was quite limited. As familiarity developed with the new system operators improved and reduced errors The operators had completed a learning process. The authors concluded with these future recommendations: increase the spatial resolution to 1024 pixels X 852 pixels and; increase the sensitivity of scanning to allow the use of 512 or even 1024 gray levels (Jackson, Dickson, and Birnie 1985). The increased spatial resolution would store more detail information in the image, while the increase in gray levels would allow better definition of structures peripheral to a landmark, especially in areas of low contrast It is interesting that the overall quality of the stored images were judged as being superior to the original

radiograph by the majority of the judges. In other dental research, all the judging groups rated the stored digital images to be of diagnostic quality (Southard 1985).

Standard video cameras offer a fast and inexpensive method for digitizing X ray films. However, as a group they lack dynamic range and spatial resolution In an average cephalometric film, the optical film density varies from 0.3 to 3.0. This corresponds to a dynamic range of brightness of almost 1000 The quality of a digitized image, via a video camera, is dependent on the transfer characteristics of the image acquisition system. This is the modulation transfer function (MTF) of the video system and on the relation between digitized pixel value and optical film density (Doler et. al 1991) As early as 1984 medical radiologists were using high resolution video cameras that were able to store radiographs at a resolution of 1024 X 1024 pixels at 8 bits per pixel (Cocklin et. al 1984). However superior results are obtained by using scanning devices for digitization. Scanning devices included: a digital camera (solid state camera); laser scanner or microdensitometer, and a drum scanner (Doler et. al. 1991).

One investigation using video capture and various image processing techniques found that the process of identifying landmarks was comparable to using a light box and digitizing tablet Landmarks in areas of high optical density, for example Nasion, are an exception. The poorer resolution here is related to the limited dynamic range of the video acquisition system. This limitation is present in all 8 bit 'frame grabbers' (Doler et. al 1991)

Laser Scanning

Laser scanning involves a rotating polygonal mirror system, used to guide a collimated low power (5 mW) laser beam, usually helium-neon, to scan across a line of the radiograph in a light tight environment The radiograph is advanced and the scan

is repeated for the second line, and so forth. The optical density of the film is measured from the transmission of the laser through a small area (for example 175 X 175 um) of the radiograph and converted to an electric signal using a photomultiplier tube and a logarithmic amplifier The electrical signal is sent to a digital chain where it is digitized to 8 bits (or up to 12 bits) by an analog to digital converter The data are then sent to a computer interface which is connected to a host computer where a storage device is provided for the digital image (Huang 1987).

The laser scanning beam passes through each pixel and the amount of transmitted light is detected. This is digitized and translated into a shade of gray in the final image. This method is cumbersome as a conventional radiograph must first be obtained. In addition it is still limited by the narrow dynamic range of the photographic film (Eppley and Sadove 1991). Some investigators decided to take a routine film, laser scan the image, digitally enhance the image, and then convert this digital image back to a regular film. However, this exposed film is not the original radiograph, but an image processed version (Fujita et al. 1987). Certainly this procedure is of academic interest only. With the quality of current image displays there seems to be no indication to step backwards and convert a digital image back to a silver halide cellulose version.

Conventional systems for exposure of Cephalograms

Conventional systems for exposure of lateral cephalometric radiographs consist of film screen combinations in which phosphors are embedded in the intensifying screen. Upon exposure, the phosphors convert radiation energy to light energy which exposes the film The rare earth phosphors were introduced in 1972 to reduce radiation exposure to the patient (Bunel and Schow 1989). This was possible because they converted radiation energy to light more efficiently, and hence had more speed.

However, there was some loss in detail (Kaugars and Fatouros 1982). One of the problems with conventional cephalometric radiography has been to visualize both hard and soft tissue components of the skull. In early attempts to better delineate the soft tissue profile, some clinicians placed barium paste down the midline of the patients face prior to exposure of the radiograph (Kaugars and Fatouros 1982). Aluminum blocks have also been placed on film cassettes to enhance the soft tissue (Freedman and Matteson 1976, and Tyndall et. al. 1987). Aluminum wedge filters have also been placed on the collimator (Freedman and Matteson 1976) Intensifying screens have also been placed with a special dye heavier in the portion of the film that will record soft tissue profile, however this system does not decrease patient radiation dose (Bunel and Schow 1989). Some investigators have examined the differences in analyzing cephalometric radiographs with and without soft tissue enhancement Surprisingly, there was no statistical difference between measurements made on each type of radiograph, even though the soft tissue outline was much clearer in the enhanced films (Bunel and Schow 1989).

Direct Capture X ray Unit

Eighty percent of conventional diagnostic radiography procedures still rely on a film screen detector system as the image receptor. The output of the procedure is a X ray film. This method of acquiring an image has been used for many years The disadvantage of using film is its cost, and the difficulty of archiving and storage. Imaging plate technology or computed radiography, introduced in the early 1980's, provides a direct digital radiograph from a conventional X-ray procedure This technology consists of two components: a reusable imaging plate for storing the latent X-ray image and a scanning mechanism to retrieve the latent image form the plate to form a digital image (Huang 1987).

The physical size of the imaging plate is similar to that of a conventional radiographic screen; it consists of a supported coat with a photo-stimulable phosphorous layer made of BaFX. Eu(X= Cl, Br, I), europium activated barium fluorohalide compounds. The photo stimulable phosphorous crystal is able to store part of the absorbed X-ray energy in a quasi stable state; thus a latent image is formed in a imaging plate just as in a film screen recorder. Stimulation of the exposed plate by a 633 nm wavelength helium-neon laser beam leads to emission of luminescence radiation corresponding to the incident X-ray energy. Luminescence radiation is collected through a focus lens and light guide into a photomultiplier tube which converts it into a electrical signal. The electric signals are then converted into digital information and form a digital image. The size of the imaging plate can be 8 X 10, 10 X 12, 14X 14 or 14 X 17 inches (Huang 1987).

One of the major advantages of the computed radiography system compared to the conventional film/screen receptor is that the imaging plate has a linear response and a large dynamic range between the X-ray exposure and light intensity emitted by the stimulated phosphor. The image scanning mechanism can automatically adjust the amount of X-ray exposure received by the plate. As a result, slight under or over exposure during an X-ray procedure does not affect the quality of the image. Hence computed radiography is capable of producing images that are comparable or superior to those of the conventional film / screen combination (Huang 1987, Eppley and Sadove 1991) Dentistry has a commercial system for conventional intraoral radiography to be performed digitally (Mouyen et. al 1989). Other dental investigators have described the replacement of a film in panoramic radiography with an x-ray sensitive scanning linear array detector with subsequent image reformation and display on a high resolution monitor. Data for successive columns of pixels are

acquired while the x-ray source and detector rotate around the head of the patient The pixel columns are assembled to present a panoramic view of a curved plane approximating the patients mandible (McDavid, Dove, Welander, and Tronje 1991)

The system is based on a solid state x-ray sensor composed of a linear array of silicon photodiodes covered by a scintillator that detects x-rays passing through the object being examined. The sensor has 512 elements with a 0.225 mm pitch and 0.5 mm effective width. A group of multiplexers successively reads out the signal from each element of the detector. An electronic preprocessor then corrects the analog offset and gain on each channel and digitizes the signal. A programmable timing circuit provides the external trigger for detector readout The detector output is in the form of 8-bit digital image data, which are transferred to a high speed programmable interface integrated with a high resolution display. Final output display is 512 X 1024 X 8 bits, corresponding to a panoramic radiograph 115.2 mm X 230.4 mm (McDavid, Dove, Welander, and Tronje 1991)

The assignment of the gray levels is based on the tonal intensity during the offset and gain calibration. The offset calibration is made in the absence of x-rays, and a value of 255 is assigned to the baseline level. The gain calibration is performed with the x-ray, and a value of 0 is assigned to the intensity of the beam during the calibration exposure. During the subsequent acquisition of the x-ray image, transmitted intensity values are graded between 0 (black) and 255 (white) with a programmed response curve . By adjusting the intensity level used for gain calibration and the shape of the response curve, the contrast of the resulting image can be manipulated and optimized. After image acquisition the density and contrast can further be manipulated by digital image processing and by adjustment of the contrast

and brightness controls on the display monitor (McDavid, Dove, Welander, and Tronje 1991).

Microdensitometer (Drum Scanner)

A drum scanner has been the traditional method of converting X-ray film to a high density and spatial resolution digital image. The disadvantages with this device are the slow scan speed and the difficulties of aligning the X-ray film in the drum. The pixel size attainable is adjustable from 25 um to 400 um, and the matrix size is variable (Huang 1987). Investigators as early as 1984 were utilizing a microdensitometer to scan radiographic films into a 1000 X 1000 matrix of film density values (Oka SW and Trussell HJ 1978). In 1984 some medical investigators were capturing chest radiographs at a resolution of 1024 X 1024 pixels at 16 bits of gray per pixel utilizing a microdensitometer (Cocklin et. al 1984). It was concluded almost a decade ago that radiographs converted into a digital format have inherently better image quality than film radiographs (Jackson PH, Dickson GC ,Birnie, DJ 1985).

Solid State Camera

Some investigators have used a digital camera to capture cephalometric films. One system described used a digital camera with a resolution of 512 X 512 pixels, following the experimental design of Jackson (1985). The investigators found they obtained an accuracy of 0.23 mm to 0.33 mm per pixel They concluded from their pilot study that computer vision equipment can provide reliable and accurate methods for obtaining orthodontic data It is interesting to note that direct digitization of the radiograph was the most reliable method used, even more reliable than using the enhanced digitally captured films They suggest that being able to change the cursor color may prevent the loss of the cursor in certain areas of the cephalometric image (Davis and Mackay 1991) A digital camera has the potential to become a very good

350 dpi. This scanner was developed at the IBM Hawthorne Laboratory to scan the art work of Andrew Wyatt.

From the TDI Proscanner the digital information is passed into an IBM Model 95 computer. From the model 95, the information is passed into another station where image post-processing is performed, and the information is stored on an optical jukebox. Alternatively, the data can also be stored on digital audio tape All of the image information is also being sent to a IBM ES 9000 mainframe computer The radiographic film being scanned is captured in a full 3000 x 4000 x 12 bits of gray That raw data is considered to be the archive quality image The definition of an archive image is a faithful (as faithful as we can technologically achieve) rendition of the original source material. The archived image is not processed in any way, nor is it enhanced This archive image represents the fundamental piece of information that should be stored, and requires 18 MB of electronic storage space. The scanning time for a 8 X 10 inch lateral cephalometric radiograph is approximately 5 minutes.

One of the unique features of this scanner is that it actually scans each line 32 times, and averages it, so the signal to noise ratio of this particular scanner is very high Consequently, one can have faith that the image off the scanner is actually very accurate. Another unusual, but extremely useful property of this scanner is that it is able to be calibrated. A process occurs prior to any scanning where a white matte and a black matte are scanned, and the response curve of scanner at the time is recorded. From that information a number of coefficients are derived, which are stored with the image. Given that, it is possible to map back from the archive accurately onto any monitor to display the image There is computer software code to calibrate the scanner and code on the delivery end to optimize the image for a particular monitor An image seen in California on an Apple computer monitor is as close to accurate as

film digitizer, but currently it still lacks the resolution and consistency required for routine clinical usage In 1987 the resolution of digital cameras ranged from 3.7 um to 176 um for pixel size (depends on the distance between the film and camera) and with matrix sizes (Spatial resolution) around 1728 pixels X 2048 pixels (Huang 1987) Recently, Hasselbad released a high resolution digital camera capable of recording spatial resolution to 2048 pixels X 2048 pixels at 14 bits per pixel per color, or 16,384 gray levels. This is quite an impressive system and one which could reproduce extended contrast range The retail market is currently awash with many digital cameras, and the future will likely see a vast proliferation of this media.

IBM Proscanner

The IBM TDI Proscanner, or a timed delayed integration scanner, is the specialized scanner utilized in this study. There have only been four of these scanners constructed so far This scanner was built to be used in conjunction with the IBM PS-2 computer system. The initial task for this scanner was to: store and retrieve text; store and retrieve pictures; permit archival of publications and quality images of art; and permit electronic distribution of archived valuable works of art It can be used with backlight or reflective illumination, and is calibrated precisely to reproduce exact color rendition (Mintzer and Delp 1990).

IBM built the PS-2 based system to scan, process, display, and print color images of art with faithful color. The IBM Proscanner is also known as the TDI PRO scanner Among its attributes are the ability to produce 12 bits of image intensity data per color at a spatial resolution of 4000 pixels X 3000 pixels. In other words, the scanner is a 36 bit color scanner or 12 bit gray scale scanner. It is possible to scan at reduced bit scans. That is approximately 300-400 dpi (dots per inch). For cephalometric radiographic films that were 8" X 10", this equates to approximately

one seen in Cleveland on an IBM monitor, and one seen on a Sun monitor located on the east coast A great deal of work is performed in capturing data and calibrating the delivery of that information.

The next step in image capture is the post-processing phase. In the post processing phase, the raw archive data is processed into a working image. The working image is also called a display ready image. This means that an image can be displayed on a particular terminal. At the workstation both the working and the archive image can be stored on digital audio tape, or on optical disk

Image Viewing

Cathode Ray Tubes (CRT)

The CRT was first developed by Braun in 1897, it was improved by Dufour in 1914 in the form of a measuring device. CRT were developed between 1930 and 1940 for radar and television, they were used as graphical output devices from 1950, and in 1960 were used for simulation of space flight. At the time these devices were very expensive and were used primarily in industrial and commercial operations Until 1970 the high cost along with the limited capabilities of CRT discouraged their use for the display of electronic graphics. As the amount of information which can be displayed by a direct beam is directly connected to the speed of amplification and deflections, and since the cost of fast memory made it difficult to preserve image lines, tubes without memory were primarily used (Brett 1991). In recent years the cost of high resolution CRT monitors has significantly decreased, to the point that most home computer systems come standard with such a good monitor that only a few years ago would have only been seen in space technology.

Viewing of radiographs requires a monitor to display at least 1024 X 1024 pixels at 8 bits per pixel (for gray scale) and 24 bits per pixel (for color) (Cocklin et

al. 1984). To decrease the memory requirements of CRT, most utilize some kind of programmable video look up table (VLUT). Depending on the type and number of CRTs used, the same image can be displayed by using different gray scale transformation tables via the VLUT, and each can be simultaneously displayed on different monitors. Each VLUT can be loaded with a number of different 8 bit gray scale transformation tables, each of which can simultaneously operate on different parts of the image display memory. Some investigators have recommended the viewing of a positive x-ray image on the monitor, rather than the usual viewing of a negative image. The viewing of the positive image was found to be less tiring, and they could switch back and forth between positive and negative images easily (Jackson et. al 1985)

Digital Processing of Cephalometric Radiographs

Recent developments in personal computer based image processing systems have enabled digital image enhancement techniques to improve the quality of digitized lateral cephalometric radiographs, and to facilitate landmark identification The approaches to digital enhancement of films may be divided into point and local operations. In point operations the output pixel value is dependent only on the value of the corresponding input pixel. Contrast enhancement, contrast stretching, and gray scale manipulation are examples of point operations. In local operations, the output pixel is dependent on the pixel values in the vicinity of the corresponding input pixel. Noise smoothing and edge enhancement are examples of local operations (Doler et. al. 1991). Image transformation does not have to be carried out on the entire image, only a part of the image may be selected if desired. By selecting only a part of the image, the time taken for the transformation is considerably reduced By having multiple

display units and several VLUT generators, it is possible to view the same image under different transformations (Cocklin et. al. 1984).

Image Contrast and Brightness

Altering the image contrast and brightness is a simple technique and is useful for compensating for different exposures. This enables a comparative assessment of two radiographs. Altering the image contrast and brightness can also be used to expand the contrast range selectively for a given small region of the radiograph so that detail within this can be more easily perceived. The authors found these adjustments to be best performed interactively (Jackson, Dickson, and Birnie 1985)

Gray Level Transformation

The gray levels f(i,j) (i,j=1,....,512) in the original image are mapped by the transformation function T to yield the gray levels g(i,j) of the enhanced image g(i,j)=T[f(i,j)]. This transformation of the gray level is done by application of look up tables. To enhance image brightness in regions of high optical density, a transfer characteristic is necessary which expands the contrast of the small values of f(i,j) and compress the large values A useful mapping is represented by a logarithmic transfer curve, g(i,j) = ln[f(i,j)] With this transformation the displayed luminance is linearly related to optical film density. Another approach to brightness enhancement consists of specifying the transformation function interactively under visual control In most cases, this yields the best result (Doler et. al. 1991)

The single most useful of all image processing techniques is gray level transformation. This is because of its speed and its ability to display image information in the most suitable way for the eye. Similar transformations have been used by radiologists as the window width and level function on CT scanners (Cocklin et. al. 1984)

Gradient Operators

Gradient operators produce image differentials and as such are noise enhancing processes. Examples of gradient operators include Roberts, Sobel and Kirsch operators. To reduce noise, gradient operators incorporate some form of smoothing Smoothing reduces resolution and there is therefore a trade-off between loss of resolution due to smoothing and loss of resolution due to noise enhancement The differences between standard gradient operators lie in the amount and type of smoothing performed. The best of these for the cephalometric radiographs appeared to be the Sobel operator (Jackson et. al. 1985).

Sobel Operator

This carries out the gradient operator in two phases· 1. Vertical smoothing and horizontal gradient extraction; 2 Horizontal smoothing and vertical gradient extraction. The results of these two passes are then combined together This elegant method employs smoothing in one direction to reduce noise while detecting edges in the orthogonal direction which are hence little affected by the smoothing. This operator was found to be most effective in testing the location of all landmarks used (Jackson et. al. 1985).

Median Filter

During the study an even better operator was discovered. It uses a median filter for noise reduction followed by a conventional horizontal/vertical difference. This was found to have improved fine detail with good noise suppression. The gradient operator really highlighted soft tissue points (Jackson et. al. 1985).

The Tracing Enhancement

Further use was made of the gradient enhancement by combining it with the original image and this is believed to be the first use of such combination images. The technique was to subtract the gradient image from the original producing dark lines where existing edges were. This was called tracing enhancement because of its similarity to the manual tracing process. As well as delineating existing edges, the tracing enhancement was good at improving the visibility of low contrast edges (Jackson P, Dickson G, Birnie D 1985).

Histogram Equalization

Histogram Equalization is a method of gray level transformation which is based on the gray level distribution in the image After calculation of the gray level histogram of the input image, the transformation function T, which is defined as proportional to the sum distribution of the gray levels, is calculated by adding up the histogram values. With this transformation, densely populated regions of gray level distribution lead to a large slope of T and thus the contrast is stretched. In sparsely populated regions of the gray scale the contrast is compressed Since contrast enhancement by histogram equalization is based on the statistics of the entire image, some gray levels will be used to display diagnostically unimportant parts of the image, such as background. To overcome the limitations of this global technique, adaptive histogram equalization (AHE) can be used (Zimmerman et. al 1988). For each pixel in the image, a region centered around the pixel is assigned. The gray values in that region are used to calculate a histogram equalization mapping which is then applied to the corresponding pixel of the input image If no special hardware is available, the algorithm for calculation of AHE is very time consuming Since brightness of a processed pixel is dependent upon local properties of the corresponding input pixels,

the image content may be altered by choosing smaller window sizes. For this reason, global histogram equalization is easier to use for contrast enhancement (Doler et al 1991). Others have found the overall result of histogram equalization to be displeasing to the radiologist (Kruger 1988)

Adaptive Histogram Equalization

Adaptive Histogram Equalization can be thought of as a hybrid processing technique that combines aspects of spatial filtering with histogram equalization. This technique attempts to optimize both techniques, with the aim being improved image quality However, as yet no significant improvement in disease detection has yet to be determined (Kruger 1988).

Image Filtering

Image enhancement by spatial filtering is a powerful method for improving image quality for human viewing. Psychophysical experiments indicate that a visual signal with accentuated edges is often subjectively estimated as more informative than the original image For this reason, filtering techniques can be used to increase the visibility of low contrast edges and to improve perception of detail (Hall 1979, Pratt 1978) The following methods are suitable for enhancing contours in digital radiological images

High Pass Filtering (Unsharp Masking)

The rationale for this type of image processing is that variations in density within a radiograph that occur over large distances are unimportant to the diagnostic task. In its simplest implementation, a deliberately blurred replica of the image is created and subtracted from the original image. When unsharp masking is applied to an image, the lower spatial frequencies are suppressed and the higher spatial

frequencies are allowed to be contrast enhanced without saturating the image display. Carried to an extreme, edge enhancement results (Kruger 1988)

The image f(i,j) is convolved with a high pass mask h(i,j) g(i,j) = f(i,j) **h(i,j). The operator ** denotes two dimensional convolution The frequency response of h is determined by the size of the filter mask and the coefficients After high pass filtering, a gray level rescaling is necessary to spread these levels over the initial dynamic range of the input image. Since an enhancement filter also processes noise, some considerations about noise smoothing are necessary (Doler et. al 1991). Using 3 X 3, 3 X 5, and 7 X 7 convolutions on 8 bit images, Jackson found these filters not to be of great benefit (Jackson et. al. 1985).

Unsharp masking algorithms are implemented in clinical Digital Luminescence Radiography (DLR) for contour enhancement. In the first stage of the algorithm an unsharp mask $f_l(i,j)$ is computed by low pass filtering the original image f(i,j). Then $f_l(i,j)$ is subtracted from the original. The result of this operation corresponds to a high pass filtered image. The enhanced image g(i,j) is obtained by amplifying the high frequency image with an emphasis parameter c, and adding the result to the original image:

$g(i,j) = f(i,j) + c[\ f(i,j) - f_l(i,j)]$, where g(i,j) is dependent on the emphasis parameter c and the kernel size of the low pass filter (Prokop et al. 1989).

Noise Smoothing in Digitized Images

A limitation of edge sharpening methods is the amplification of high spatial frequency noise, which arises from the image acquisition system. Noise effects can be minimized by using appropriate smoothing techniques (Doler et. al 1991).

Image Averaging (Statistical Differencing)

If an image g(i,j) is formed by averaging N different noisy images, the signal to noise ratio (SNR) is proportional to the square root of the numbers of images averaged: $SNR \sim \sqrt{N}$. Image averaging is a very efficient method for noise reduction. Many image processing systems with real time features perform this operation within a few seconds (Hall 1979). Image averaging uses the local standard deviation values and this characterizes the local contrast Each local part of the image can be set to a constant average brightness and areas of low contrast can be enhanced to improve their visibility An attractive feature of this technique is the control the user has in avoiding enhancement of edges that are already well defined by careful use of the standard deviation gain function. In unsharp masking these edges are often over enhanced to the extent that the density values on either side of the edge reach maximum and minimum density values thereby losing information. This is often called edge ringing. This can be avoided in statistical differencing because those edges that were originally well defined (and therefore have high contrast) can be left unchanged, while enhancing areas of lower contrast (Cocklin et. al. 1984). The statistical difference filter in the form described by Pratt 1978, is an adaptive high pass filter, and was found to be superior to the basic high pass filter and produced improved visualization of the maxilla in particular (very helpful in determining palatal plane) (Jackson PH, Dickson GC ,Birnie, DJ 1985).

Edge Preserving Smoothing

A simple linear filtering method for noise reduction is the moving average filter where each pixel of an image is placed by the average of the surrounding pixels. A basic difficulty with this is or similar smoothing operations is that they also blur edges and contours in the image. The conflict between noise reduction and edge degradation

may be solved by using edge preserving smoothing operations. A simple algorithm performs a smoothing operation only in the homogenous regions of the image (Hall 1979). A more sophisticated procedure for edge preserving uses pentagonal and hexagonal masks for computing local averages (Nagao and Matsuyama 1979). This smoothing method also has the ability to sharpen blurred edges (Doler et. al 1991).

Advantages of image processing in cephalometrics

The advantages of utilizing digital image processing in cephalometric radiography include:

1. Reduction in patient dose;
2 Reduced processing costs,
3. Improved storage of and access to information;
4 Manipulation of the original image to improve quality and readability;
5. Selectively process parts of the image and improve landmark clarity and location;
6. Direct digitization of the original or enhanced image from video monitors (Jackson et. al. 1985).

An underexposed cephalogram which is inadequate for orthodontic use, can be enhanced to actually show more details than the original film (Doler et al 1991). However others found the best enhancement of a particular point depended on the quality of the initial film (Jackson et. al. 1985). Being able to enhance poor films is extremely important in clinical orthodontics. Poor cephalograms will be able to be utilized, preventing excess patient radiation exposure from retaking films. Extrapolating on this idea, image enhancement techniques may be applied to routine cephalograms. These cephalograms would be taken using lower exposure times, and the resultant underexposed film, may be able to be enhanced and become a useful diagnostic aid. This process would result in a much lower radiation dose to the patient.

The specific techniques required for this to take place are common algorithms in most image processing software Histogram equalization (or gray level transforms) is one of these commonly available algorithms, and is quite useful for contrast enhancement of poor quality cephalometric films. Another common algorithm is one for filtering edges. The enhanced image with accentuated edges subjectively appears more informative than the unprocessed image. This is because the relevant anatomical structures are more visible (Doler et. al. 1991).

One of the problems encountered in using video acquisition systems was localization of landmarks in areas of high optical density, for example, the nasal bone. This problem is related to the limited dynamic range (256 gray shades) of the video system, and is present in all 8 bit video frame grabbers. One solution of this problem is to combine two images taken at different brightness levels (Kundel and Mezrich 1987). Noise reduction filters are then used to enhance the region of interest (Doler et. al. 1991).

Accuracy of Cephalometric Measurements

Accuracy of cephalometric measurement is also determined by the spatial sampling properties of the image acquisition system The resolution of details in a digital video system depends on the size of the image matrix and the modulation transfer function. For a radiograph of 250 mm side length and a digital image matrix of 512 X 512 pixel, the maximal achievable resolution is about 1 pixel/mm. In addition, the modulation transfer function of the video camera used decreases rapidly at higher spatial frequencies. Thus only a lower resolution can be obtained for the given image size This effect may partly be compensated by high pass filtering of the digitized image To increase the spatial resolution by a factor of 2, a 1024 X 1024 image matrix is necessary. Since the resolution of a video system is dependent upon

the bandwidth of the video signal, this image requires a high resolution TV system (Doler et al. 1991) This high resolution is defined as HDTV (High Definition Television), and has a spatial resolution of 1024 pixels X 1536 pixels.

Reliability of Head Film Measurements

The science of cephalometry is subject to numerous types of error. Regardless of the use to which head films are used, it is important to understand the amount of error involved (Baumrind and Frantz 1971). It seems to be quite common for orthodontists to regard the method error in the compilation of data from cephalometric headplates as a matter of minor significance. For small linear and angular variations the biological variation can be so small that the variance registered in a cephalometric study may be strongly influenced and even dominated by the method error This effect is further enhanced if such parameters constitute the difference between two measurements because the variance will then be influenced by the method error in two registrations. For instance, this will be the case for growth changes and for treatment changes calculated from measurements on radiographs taken on different occasions. It is imperative that the method error be reduced to an absolute minimum (Bondevik et. al. 1981) If sufficient care is not taken, these errors can be of such a magnitude that they can pervert the result in single individual analyses. In group studies these errors will have less significance, yet Slagsvold (1969) found that a misreading in one out of 53 observations can change the distribution of the variable from normal to one of extreme skewness There are four main types of error, Projection error, Landmark Identification error, Tracing Error, and Measurement error.

Projection Error

Projection error occurs in the conversion of a three dimensional skull into a two dimensional radiograph Projection error varies with relative positions of the film, the x-ray tube and the patient Since the X-rays which produce the shadow are non parallel and originate from a very small source, head films are always distorted enlargements. The enlargement factor varies depending on the plane at which the estimated point lies. Lateral cephalograms are further distorted by foreshortening of distances between points lying in different planes, and by radial displacement of all points and structures not on the principal axis (i.e. central ray) (Baumrind and Frantz 1971) Many investigators have discussed projection errors in cephalometry (Adams 1940, Brodie 1949, Thurrow 1951, Franklin 1952, Kean 1958, Hallet 1959, Hixon 1960, Bjork 1962, Salzmann 1964, and Baumrind and Frantz 1971), and some have suggested methods of reducing errors of this type (Adams 1940, Wylie and Elsasser 1948, and Vogel 1967) However, most remedies are quite cumbersome, and this has restricted their routine use.

Landmark Identification

Many investigators have shown landmark identification to be an important source of error in cephalometry (Bjork 1947, Hixon 1956, Hatton and Grainger 1958, Miller et. al 1966, Savara et. al 1966, Richardson 1966, Carlsson 1967, Baumrind and Frantz 1971, Midtgard et. al. 1974, Broch et. al. 1981, and Bondevik et. al. 1981). In a classic study using computer technology, landmark identification errors were shown to be specific to each landmark and show a characteristic envelope of error (Baumrind and Frantz 1971). These authors concluded:

1 Even when one is replicating assessments of the same head film, errors in landmark identification are too great to be ignored,
2. The magnitude of the error varies greatly from landmark to landmark;
3. The distribution of errors for most landmarks is not random but is rather

systematic in the sense that each landmark has its own characteristic and usually non circular envelope of error;
4. The perceptual task involved in identifying landmarks varies from point to point (Baumrind and Frantz 1971).

Most often the judge is asked to estimate the position of a point on an edge. The precision with which this takes place is largely a function of how sharply the edge folds in the region of the point being estimated. Where the edge folds very sharply, (as at the incisor edge) the estimates are very good. Where the edge is a gradual curve (e.g. A or B pt or Go) the task is more difficult and the errors tend to be proportionately larger, and also to be distributed along the edge itself (that is along the edge of the skull). This holds true for Porion and Menton, both of which proved more variable than expected. A corollary factor is the sharpness of the viewed edge, this is the degree to which the edge contrasts with the surrounding area. Most of the points orthodontists locate lie on the surfaces of the skull. Those structures which lie within the confines of the skull have a greater likelihood of being confounded by noise from adjacent or superimposed structures Some landmarks are confounded by so much noise that the observer is often asked to identify a point for which there is frequently no direct physical evidence at all on the head film When locating the lower incisor apex, the judge is forced to make a conceptual operation based upon his general knowledge of how long a root is, its expected rate of taper, given only the perceived conformation of the crown and visible portion of the root. In making such perceptual judgments, the prior experience of a judge is an important factor, and it is in the judgments of this sort that an experienced operator tends to be more reliable than a novice (Baumrind and Frantz 1971)

Several observers have referred to the particularly low level of tracing reliability of the incisor teeth (Hixon 1956, Backlund 1963, Mills 1964, 1966, Bennet and Smales 1969, Kvam and Krogstad 1969). A further problem appears to be rigor

of definition of landmarks. Differences of opinion commonly relate to locating landmarks in different places, purely on the understanding of the given definition for a given landmark. Some investigators have even redefined landmarks in an attempt to reduce confusion when locating landmarks (Baumrind and Frantz 1971). The accuracy of landmark recording is also dependent on: the characteristics of cranial structures; the general quality of the headplate; blurring of the anatomical structures caused by secondary radiation or movement during exposure; precision of the recording method; and the accuracy of the operator (Broch J, O Slagsvold, and M Rosler 1981). These authors considered the headplates used were of a high quality, and that the recording method had a high degree of precision. Their conclusion was that the characteristics of the cranial structures are responsible for most of the variations in reproducibility of landmarks and also for the way in which the registrations are spread along the two axes (Broch J, O Slagsvold, and M Rosler 1981). Other investigators believe that because the greatest number of errors occur in landmark identification, efforts to improve accuracy in landmark definition should be directed primarily toward improvement in image quality (Eppley and Sadove 1991).

Sharpness of a Cephalometric radiograph

Other investigators have given more importance to unsharpness of the image and lack of contrast in the radiograph in selecting landmarks (Cohen 1984). Sharpness is defined as the rate of change of image density across the boundary between a point of interest and the surrounding area. Radiographic image sharpness depends on a number of factors: geometric, photographic, sharpness of the subject itself, and movement that may occur during exposure. The anode, or focus of the X-ray machine is not a point, but has an appreciable area. This has the effect of producing penumbra on the X-ray film. A number of photographic considerations are significant, and

should be taken into account when examining cephalometric films. A radiographic film has a certain grain size, and while this does set a theoretical limit on resolution, in practical terms the grain size is so small as to become unimportant as compared with other causes of unsharpness. Intensifying screens used to reduce the X-ray exposure, produce a loss of sharpness because the light from them tends to diffuse. Also, the use of intensifying screens contributes considerably to a very coarse graininess of the image, known as quantum mottle. Quantum mottle occurs because the screen amplifies effective sensitivity to the point at which areas of fill are subjected to a small X-ray exposure and produces a patchy latent image. This can be compared to the effect of a light shower of rain on pavement which gives a patchy wetting effect, while a heavier shower produces a more even appearance. Unsharpness of the subject itself leads to an unsharp image and similarly movement during the exposure will also produce unsharpness (Cohen 1984). However Halse (1978) found poor quality images made little difference in the accuracy of locating landmarks

Contrast

Contrast is defined as the difference in density between adjacent areas. Factors in the lack of radiographic contrast include those due to the object, the image recording system, and the conditions of viewing. The kV level is important because high kV values tend to level out differences in relative X-ray absorption and reduce the contrast. The presence of scattered radiation reduces contrast by detracting from the difference in X-ray exposure of adjacent parts of the image (Cohen 1984).

Electronic Plotting of Cephalograms

In the past few years cephalometric points have been commonly recorded on a data processing (digitizing) tablet This input device is very precise, commonly with an accuracy of approximately 0 01 mm Many of the more modern tablets have an

accuracy in excess of 0.001 mm. The data processing tablet is the preferred instrument of architects and industrial draftsmen. Its sole disadvantage is the dissociation between the drawing area, the surface of the tablet and the area of visualization, the screen Electronic plotting of lateral cephalograms can be employed in two ways: Direct digitization of points from the lateral cephalogram placed on a illuminated digitizing tablet and; Digitization of points of a cephalometric tracing placed on the digitizing tablet. Richardson (1981) made a comparative study of the reproducibility of the use of an electronic digitizing tablet when compared with a standard tracing. He found the digitizer to be only marginally superior in the analysis of lateral cephalometric films Theoretically, as the digitizer is a highly accurate measuring instrument, one might have expected a considerable gain in accuracy from the use of the instrument when compared with the measurements made with a ruler and a protractor. Richardson 1981 pointed out that much of the measuring accuracy of the digitizer is lost when it is used on poorly defined points on an X-ray, whose location is often subjective. More accurate measurement between poorly defined points offers little gain in precision of results.

Direct Digitization of Cephalograms

A study with surprising results reported that tracing a film prior to digitizing was less reliable than directly digitizing the landmarks. They surmised that measurement of a tracing required estimates. First, drawing the anatomic landmark; and second attempting to record the exact position of the pencil line, which has an appreciable width (Cohen 1984). However, one must bear in mind that this study tested reproducibility of measurements, not reliability of measurements.

Landmark Location from a Computer Monitor

Another study tested landmark identification by directly locating landmarks on a computer screen Unfortunately this was a preliminary report, with no results stated (Baker et. al. 1990).

Tracing Error

Tracing error is caused by lack of clarity of cephalometric landmarks due to superimposition of structures, blurring of the image due to movement during exposure, lack of film contrast and emulsion grain. Measurement error associated with the thickness of the pencil line and the perceptive limits of the human eye also contribute to tracing error (Gravely and Benzies 1974). Tracing error can be the major source of systematic error (Backlund 1963, and Carlsson 1967). If these errors are random they can be neutralized in group studies, i.e using a large sample with multiple tracings of each individual film (Gravely and Benzies 1974) Inter tracer error is generally higher than intra tracer error (Hixon 1956, Broadway et. al. 1962), and the level of error is affected by experience and training (Kvam and Krogstad 1969).

Measurement Error

May arise from incorrect sequence of digitization, or from movement of the headplate during the registration of the lateral cephalometric films.

Analysis of Cephalograms

The same angle measured on two separate occasions may represent the reliability of repeated measurements, that is:

$$SD = \sqrt{\frac{E(X_1 - X_2)^2}{n}}$$

where X_1 = first measurement of the cephalogram, X_2 = time 2 measurement on cephalogram, SD = standard deviation, and n= number of Cephalograms traced.

From the population SD, an unbiased estimate of the SD of individual tracings σ, has been made from the relationship

$s = \sqrt{2\sigma}$, Thus $\sigma = \sqrt{\frac{E(x_1 - x_2)^2}{2n}}$

The value of σ,is a measure of the accuracy with which a particular point can be traced and measured by a particular tracer. Thus the confidence limits within which that particular tracer can work have been established for each of the three tracers as X ± tσ, where X is the observed measurement. the value of t is dependent on the level of Probability adopted.

Confidence Limits

$$X_1 - X_2 \pm t\sqrt{2\sigma = (X_1 - X_2)} \pm t\sqrt{\frac{E(X_1 - X_2)}{n}} \text{X1-X2}$$

The replication of tracings only reduces the tracing error and as tracing error is reduced, other systematic errors become more important.

Computer Location of Cephalometric Landmarks

Automating the process of landmark identification on X-ray images may produce higher quality data and improve data capture productivity. An early attempt at computer landmark recognition was undertaken by Cohen, Ip, and Linney (1984). They used a Clip 4 computer and captured the image with a video acquisition system of 96 X 96 pixels at 6 bits per pixel (64 shades of gray). The image display used was a TV monitor with 0 39 mm resolution. The study compared: locating landmarks with the computer; manual landmark location using a digitizing tablet with 0.017 mm accuracy; and manual selection of points on the video monitor. They looked at finding two specific points - Me and S They then used several image processing techniques to enhance the video image. Poor results were obtained from both the video techniques, and they suggested this was probably due to the poor screen

resolution of 0 39 mm (Cohen, Ip, and Linney 1984). Currently, a monitor resolution of 0.28 mm is regarded as the norm.

A method is proposed for automating landmark location and data extraction form craniofacial X-ray images using neural networks and match filtering techniques (Baker and Baumrind 1990). When observers locate landmarks, there are systematic errors associated with each landmark and each judge. These errors are the result of the specific features of each landmark and the conceptual and perceptual definition of each judge. The systematic errors are characterized by envelopes of error for each landmark which differ in both size and shape (Baumrind and Frantz 1971). Further, when several expert judges assess the same landmark on a series of images, the assessments of different judges typically show systematic differences with respect to the group mean. Even the written definitions for conventional craniofacial radiographic landmarks vary considerably between different expert judges. Further, complications are introduced by the fact that landmarks vary in shape and gray scale characteristics from patient to patient and at different stages of growth in the same patient (Baker and Baumrind 1990).

In an interesting, comprehensive study three hundred lateral cephalometric radiographs were electronically captured with a digital camera (Baker, Liu, and Baumrind 1990). All cephalometric landmarks were known from previous work, and the investigators used neural network technology to teach the computer to find landmarks on the digitized films. Neural network methods represent a promising approach to the problem of automatic landmark location and may overcome some of the limitations of earlier methods of template matching and feature observation. They are trainable, adaptive and operate by examining many hypotheses simultaneously using parallel nonlinear elements connected by links of various weights (Baker,

Baumrind and Curry 1990). Perhaps to utilize computer recognition of landmarks, our cephalometric landmarks need to be changed to other points that can be easily selected automatically by a computer using image processing techniques (Cohen et al. 1984). It should be remembered that many of our cephalometric points currently used have come from prior days of craniometry, and historic analyses With large databases of information, it would be easy to produce a new set of computer friendly landmarks (ones easy for automatic landmark recognition), and new values associated with growth and treatment planning

Probability of Cephalometric Point Location

The probability of locating one point accurately is 0.05. For using a cephalometric analysis of 16 points , the probability of obtaining all 16 landmarks successfully is 0.05^{16} or 44%. Thus there are 56 chances out of 100 that the analysis will contain on the average at least one landmark value which deviates by more than two times the standard deviation from the true head film value for that landmark. When we consider lateral cephalographs are used in pairs for comparison, the average chance that all thirty two estimations in the pair will be located with errors of less than two standard deviations is slightly less than two chances out of 10. The vast majority of head film analyses based on a single estimate for each landmark cannot help but be flawed, the more so since we have no way of gauging which estimates are the bad ones. The obvious answer for more precision is to replicate measurements. If this is done, the Probability that the mean pair of independent estimates will be greater than two standard deviations from the true value is the Probability of erring in the same direction (or tail) from the true value in both estimates, namely 0.025 X 0.025 or 0.625%. The average likelihood of completing a 16 point replicated analysis on a single film without undetectable error becomes 98% If more than 1 replication was

used, the likelihood of undetected error would exponentially decrease (Baumrind and Frantz 1971).

Photographic Compact Disk Read Only Memory

Photo CD is the colloquial abbreviation for Photographic Compact Disc Read Only Memory. Audio Compact Discs have become common place in recent years, but only recently has the Eastman Kodak Corporation released Photo Compact Discs. Compact Discs have become a common storage media because they can hold approximately 600 MB of information In contrast, commonly used high capacity floppy disks are limited to 1 2, or 1.44 MB of information. A Photo Compact Disc ROM is a plastic disc on which digital information can be stored Photo Compact Disc ROM is an interactive, digital medium One is not reliant upon an analog linear progression of information, as is the case with a videotape A videotape in which data is stored (pictures and sound) is presented at a fixed rate, which takes you from point A to Point Z, passing all the intermediate points in the process. Digital technology does not restrict one to this A-B-C format If one wishes to skip to point C, and then back to point B it is simple and does not involve any rewinding

A Floppy Disk stores data on magnetic coated plastic (much like a cassette tape). A Compact Disc is constructed of plastic, and its storage is not magnetic phenomena. Information stored on a magnetic surface will eventually degrade, sometimes for no other reason than being played too many times A Compact Disc uses laser technology to write and read data. Once data is written to a Compact Disc, it is non-erasable. The major advantage of a Photo Compact Disc ROM is its durability. The only thing that comes into contact with the disc is the laser, hence there is no possibility of image degradation from repeated use.

A Kodak Photo Compact Disc is gold plated. Unlike most metals which begin to corrode after about ten years, gold will not corrode. Audio Compact Discs use an aluminum coating, which can oxidize over the years, making the data on the disk inaccessible Gold is a very reflective substance, which helps to optimize data exchange. The less reflective the disc's surface the more likely it is that encoding errors will occur, which constitutes an overall degradation of image quality. A Kodak Photo Compact Disc is reported to have a shelf life equal to or exceeding that of conventional photographic emulsions However, with Photo CD you can copy the information from one disk to another, and if fifteen to twenty years the shelf life will start over again In other words you would have created an entirely new original, with no image degradation from the original. The limiting factor is the physical life span of the compact disc Fingerprints and dirt on the surface of the disk will not directly affect the recorded signal, but such blemishes will reduce the brightness of the light reflected off the surface, which may result in playback omissions.

How an Optical Disk Works

This device uses a laser diode as the light source for recording and playback of data In the recording process, direct modulation by the recording (input) signal results in a focused laser beam irradiating and heating a tellurium suboxide (TeOx) coated disk at a chain of points. Because of this heating, the TeOx thin film material changes its reflectivity to a higher value. It is this change in reflectivity which stores the data Playback of stored information is achieved by detection of light reflected off the disk at these points (Yoshida 1982)

How Photo Compact Disc ROM Works

Conventional picture taken.

Currently, the film size is limited to 35 mm, however larger format films will be supported in the near future.

Conventional film processing.

The particular process corresponds to the individual film emulsion.

Scanning of the negative or slide.

A proprietary Photo Compact Disc ROM Film Scanner is used to scan each 35 mm frame The image is translated into a RGB digital image with a resolution of 2048 lines X 3072 pixels. As the Kodak film scanner is scanning the film, it interacts with the Kodak Photo Compact Disc ROM Data Manager to automatically adjust each frame according to exposure variations, lighting, and the specific type of film the image was recorded on, to obtain the best possible image quality. This process can also be manually adjusted to allow individual characterization of a particular image

Data Processing

The scanning process results in a 18 MB image file At this size only 30 individual images would fit on a single Photo Compact Disc ROM In addition, it would take a full two minutes for a PHOTO Compact Disc ROM or a Compact Disc ROM reader to load a single image A proprietary Photo Compact Disc ROM Data Manager compresses the image from 18 MB to somewhere between 4.5 MB to 6 MB. This process occurs concurrent with the scanning process. The data manager transposes each image into an IMAGE PAC FILE, and acts as a digital table of contents for the Photo Compact Disc ROM. Simultaneously, the data manager sends the images to the Kodak Photo Compact Disc ROM Writer, which writes the digital files to the Photo Compact Disc ROM. The data manager also transmits the contents

of the Overview pack to the Photo Compact Disc ROM Index Printer, which generates a single print that contains all the images stored on the disc in thumbnail format. This whole process requires an enormous amount of data processing, and would not be possible without the proprietary hardware supplied as the Photo Compact Disc ROM Data Manager

The Kodak workstation would not be possible without the automated speed and consistency of the Photo Compact Disc ROM scanner and Data Manager. Until recently, scanning required meticulous manual operation, and existing software for image control was clumsy and complicated. However the Kodak system is extremely easy to use, and produces consistent quality.

Photo Compact Disc Data

Virtually all the visual information in each original photograph is captured in the 18 MB file. This file is compressed to approximately 4 5 MB for storage on the optical disk. Each picture is stored in one file at five different resolutions (Image Pac):

1. 128 pixels X 192 pixels (lowest resolution - Base/16)
2. 384 pixels X 256 pixels (lower resolution - Base/4)
3. 768 pixels X 512 pixels (Base Image)
4. 1536 pixels X 1024 pixels (higher resolution of Base 4)
5. 3072 pixels X 2048 pixels (highest resolution of Base 16)

Hence it is possible to use the resolution that best suits your needs. Normally approximately 100 images can be stored on a Photo Compact Disc ROM However one may store up to 800 images on a Photo Compact Disc ROM at TV resolution. This requires that the lowest three resolutions be the only ones stored. Every time images are added to an existing Photo Compact Disc ROM, the overall storage is reduced. This storage loss is a consequence of separate sessions all having to add different disk directories Existing directories cannot be overwritten Consequently, to maximize disk storage, it is best to store all images on the Photo Compact Disc

ROM in one session Also, some of the older Compact Disc ROM players are unable to read multi session Compact Discs The majority of the newer Compact Disc ROM players are multi-session compatible.

Image Viewing

One of the interesting features of this system is the multitude of TV-monitor display capabilities one can access with the Photo Compact Disc system. When viewed on a TV, the image is much sharper than a regular TV picture Colors are amazingly bold, and one is not limited to just viewing the image, but one can zoom in, zoom out, rotate, and crop the image. Photo Compact Disc ROM images can be transferred to a analog medium such as videotape. Images, sound and text may easily be combined to construct a whole presentation.

Photo Compact Disc ROM can integrate with any output display from NTSC and PAL to HDTV (High Definition TV). The digital information contained in the Photo Compact Disc ROM is 16X greater than current television standards, and four times greater than the standard HDTV The base image with a luminance resolution of 768 pixels X 512 pixels is used for the NTSC and PAL television systems. The higher resolution of image components of 4 base (1024 pixels X 1536 pixels) and 16 base (2048 pixels X 3072 pixels) support HDTV, which utilize 960 or more lines, as well as the Kodak Photo Compact Disc ROM Thermal Printer which utilizes 2048 lines of resolution. The lower resolution components of 4 base and 16 base allow rapid access, which is useful for displaying more than one image simultaneously, displaying vertically captured images; and displaying or printing Overview pacs, icons and index prints. A base/16 image contains 1/16 the number of pixels contained in the base image, while a 16 base image contains 16X the number of pixels contained in the base image

Photographic Duplication

In conventional photographic duplication the second generation dupe is significantly lower in quality than the original, and the third generation copy is significantly lower in quality than the second generation dupe, and so on with further reproductions. The digital image can be duplicated ad infinitum without any loss of image quality. A Photo Compact Disc ROM image can be sent to a variety of sources for a variety of uses, and each source will be in effect working with identical images.

STATEMENT OF THESIS

The purpose of this research is to determine the spatial resolution necessary to use digital representations of lateral cephalograms in the same way as clinicians and researchers now use lateral cephalometric radiographs on silver halide cellulose. The specific aims include:

1. Subjective evaluation of various operators ability to discriminate between the digital representations of the same radiograph at three spatial resolutions shown in a A B comparison on the computer screen.
2. Objectively evaluate various operators ability to locate 25 cephalometric landmarks on conventional lateral cephalometric radiographs, and the location of these same points on digital representations of the same radiographs

MATERIALS AND METHODS

This study used twenty five lateral cephalometric radiographs chosen at random from the 428 radiographs used to develop the Bolton Standards These 25 films were photographed and transferred to a photographic compact disc. Five images were chosen randomly from the sample of 25 and used in a subjective test on spatial resolution. Using two matched monitors, a spatial resolution test was devised to compare the same radiograph at three different levels of spatial resolution An optimal spatial resolution was determined by this test.

The 25 sample images on photographic compact disc were then displayed on the computer monitor at this optimal spatial resolution and the 25 landmarks were selected for each image These points were located in a x,y coordinate system, using the same point on each film as the origin for the x,y coordinate axis system Landmark location was then compared between the optimal spatial resolution image and the original radiograph. Each original radiograph was traced, and 25 landmarks identified The landmarks consisted of 21 anatomic points, and four fiducial reference points. Each datum point was digitized into an x,y coordinate, enabling the landmark location reliability to be compared inter-examiner, and intra-examiner

Sample

Twenty five lateral cephalometric films were selected from 428 cephalograms using a random number generator from the Statistical Package for the Social Sciences (SPSS) for Windows Version 5.0 1 (SPSS Inc Chicago, Illinois) computer software.

All radiographs were part of the Bolton-Brush Growth Study. The 428 radiographs were used to develop the Bolton Standards of Dentofacial Development

All cephalograms were indexed with corner fiducials according to the method described by Baumrind (1971). The use of corner fiducials allowed mathematical stacking of all landmarks identified on the same film regardless of whether the landmark is located using an acetate tracing or with a cursor on a computer monitor The use of the corner fiducials creates a common x-y coordinate system allowing comparison of the spatial relationship of landmarks located by different examiners, by the same examiner at different times, and by examiners using different media i.e tracings and digitized photographs All of the original tracings were digitized using an x-y coordinate digitizing pad (Summagraphics ® Summasketch ™, Model MM 1201, Fairfield Connecticut) A customized computer software program, (Plotodont, written in Turbo Basic by Dr. Robert Mensforth of Cleveland State University in May 1993), was used to store the data as x-y coordinates. As suggested by Baumrind et al. the gold standard for comparison was the average of all tracings (Baumrind et. al 1971). Averaging of landmarks was accomplished by calculating the mean x coordinate, and the mean y coordinate for each of the located landmarks for each radiograph. Each individual landmark location was then plotted as a deviation from the mean value. Consequently, each landmark location was able to be identified from a mean value of 0,0 on a x,y coordinate axis system This information was transferred into a spreadsheet program (Microsoft Excel for Windows Version 4.0, Microsoft Corporation, Redmond, Washington D.C.) where initial data organization occurred. This same program was also used for statistical analysis.

Subjective Testing of Spatial Resolution

Three separate tests were conducted under the subjective testing of spatial resolution. All twenty eight judges performed each of the three tests The first test examined any bias present in the monitors used for this study The second test examined the ability of the twenty eight judges to choose the radiograph which offered the most information when shown a radiograph viewed at a low resolution (384 pixels X 256 pixels) and the same radiograph viewed at a medium resolution (768 pixels X 512 pixels) The third test examined the same judges to choose which radiograph offered the most information when shown a radiograph at medium resolution (768 pixels X 512 pixels) and the same radiograph at high resolution (1536 pixels X 1024 pixels).

Sample

A random sample of 5 films was chosen from the main sample of 25 lateral cephalometric radiographs selected from the Bolton Standards.

Judges

The judges were selected from a group of experienced orthodontists, experienced oral surgeons, orthodontic residents, and oral surgery residents from C.W.R.U.. The judges had no prior knowledge of the study, nor were they aware of any hypothesis being examined An actual copy of the judging sheet is given in appendix B.

Method

The 25 randomly selected sample of Bolton Standard lateral cephalometric films were photographed using a Canon EOS 620 35 mm camera, 100 mm Canon 1:1 Macro lens, with 100 A.S.A. Kodak Ektachrome slide film and backlight illumination. The fluorescent light box (Logan Electric Spec Mfg. Co. Model 810/920, Chicago,

Illinois) was placed on the base of a copy stand. The camera was mounted to the copy stand, and the target film distance was adjusted to allow the radiograph to fill the field of view. The camera was adjusted so that the central ray of the camera passed through the center of the radiograph at ninety degrees. A fluorescent filter (Hoya Filter # 32-4 Tokyo, Japan) was used on the camera to subdue the green hues that may result from fluorescent illumination. Three exposures were taken of every radiograph, f/8, f/11, and f/16 The shutter speed was selected electronically, utilizing the aperture priority function of the camera. A remote shutter release was used to eliminate any camera shake during the exposure. The Kodak Ektachrome film was developed by the C.W.R U Medical Photography Service in February 1993.

The processed transparencies were then inspected by Mr. Tony Cannata of A C. Color Labs The transparency exhibiting the best exposure was selected for each radiograph, and these slides were then scanned and placed onto a Kodak Photographic Compact Disc, as previously described. Each photographic slide is placed onto the compact disc at five different resolutions:

1. 192 pixels X 128 pixels;
2. 384 pixels X 256 pixels,
3 768 pixels X 512 pixels,
4. 1536 pixels X 1024 pixels,
5. 3072 pixels X 2048 pixels

Consequently each electronic image was identical except being presented at different spatial resolutions. An excellent opportunity was presented to examine the same image at different spatial resolutions The spatial resolution test is important because the amount of memory necessary to store the images is related to the pixel density by a factor of 4 Considering the raw 2048 pixel X 3072 pixel images use approximately 12 MB of disk space per image, it was apparent that electronic cephalometric images are enormous

All tests involving display of cephalograms on a computer monitor used an Apple Quadra 800 (Apple Computer Inc Cupertino, California, Model number M1206), with 72 MB of RAM, 1 GB hard drive, and a internal multisession CD Rom drive. For subjective testing two matched 17 inch monitors (SuperMatch 17-T by SuperMac Technology, Sunnyvale, California, Model number CDM-4710) were used to display the same image shown at different spatial resolutions in a side by side comparison.

Of the five available spatial resolutions stored on the Photo CD, the following three resolutions were chosen for our testing procedure:

1. 384 pixels X 256 pixels termed Resolution L (for Low),
2. 768 pixels X 512 pixels termed Resolution M (for Mid); This is TV resolution;
3. 1536 pixels X 1024 pixels termed Resolution H (for High) This is the HDTV resolution.

The lowest resolution of 192 pixels X 128 pixels was omitted from the test as preliminary tests showed this resolution was noticeably inferior to the images with higher spatial resolutions. The highest spatial resolution was not tested as preliminary tests showed the 768 X 512 and 1536 X 1024 images were likely to be sufficient for orthodontic needs.

The two monitors were matched by a calibration instrument (SuperMac Monitor Calibrator by SuperMac Technology, Sunnyvale, California, Model number CDC-35). This calibration tool adjusted the monitor's visual output to be as close as possible to each other. The virtual monitor function of the Apple McIntosh was used to enable films to be placed on either monitor at will. All the images were opened using a computer graphics program (Adobe Photoshop Version 2 1., Adobe Systems Inc., Mountain View, California), and displayed on the screen in such a manner that no information about the image was known to the individual judging the radiographic

images All the images were stored in RAM, which allowed the principal investigator to display the test images immediately, as the images were not having to be opened from the slower responding hard disc, or much slower access time CD Rom.

There were five test images at three different spatial resolutions involved in this study. The images were shown in a predetermined order on either monitor A or monitor B (see previous table for order) to prevent images being mismatched to the judges data entry form (see Appendix B) To prevent confusion on the part of the principal investigator when swapping one film from monitor A to monitor B, duplicate copies of all the images were opened There were thirty images opened in RAM for the test. During the setup for the test, one image for each of the spatial resolutions of the different radiographs were placed on Monitor A, with the identical copies being placed on Monitor B. The display of the different films at different spatial resolutions was a simple procedure, with little possibility of images not being displayed in the correct sequence To display the different resolution images at the same size it was necessary to magnify the viewing resolution of the low resolution film four times, and the medium film two times. The low, medium and high resolution films were shown at viewing ratios of 4:1, 2 1, and 1:1 respectively.

The images were shown to one judge at a time The judges were allowed to view each image pair for as long as necessary All the subjective tests were carried out in the Enlow-Gould Room of the Orthodontic Department C W.R U.. This viewing room had no windows, and the overhead lighting was turned off to prevent extraneous light from interfering with the judges viewing the computer screen The time taken for each test averaged twenty five minutes

The subjective spatial resolution test involved the judges rating the images presented to them, two at a time. One image was presented on a computer monitor

placed on the left hand side, termed Monitor A, and the second image was presented on the monitor placed on the right hand side termed Monitor B. The judges were told they were presented with two images of the same cephalometric radiograph, and were asked to determine if:

1. The information content of the images was equal;
2. More information is apparent on the image on Monitor A or;
3. More information is apparent on the image on Monitor B.

The image pairs were each shown in the following sequence

Test of each image	**Monitor A**	**Monitor B**
1	L (256 X 384)	Ll (256 X 384)
2	M (512 X 768)	Mm (512 X 768)
3	L (256 X 384)	Mm (512 X 768)
4	M (512 X 768)	Hh (1024 X 1536)
5	H (1024 X 1536)	Mm (512 X 768)

L, M, H, represents the Low, Medium and High resolution images respectively. Ll, Mm, and Hh are exact copies of L, M, and H respectively.

The test was designed to identify bias between the monitors, intra examiner reliability, and inter examiner preference The highest resolution image (C) was not directly compared to the lowest resolution image (A). The rationale behind this came from the pilot study where the mid resolution image (B) was rated as consistently better than the low resolution image A Hence, Image C could be inferred to be superior to image (A)

Monitor Bias

This test used both the low resolution (384 pixels X 256 pixels) and the medium resolution (768 pixels X 512 pixels) lateral cephalometric digital images The first test in each series examined identical images of the low resolution (384 pixels X 256 pixels) image displayed on Monitor A, and on Monitor B. The second test in each

series examined identical images of the medium resolution (768 pixels X 512 pixels) image displayed on Monitor A and an duplicate at the same spatial resolution on Monitor B.

Objective Testing of Cephalometric Landmark Location

Sample

The same twenty five original lateral cephalograms as previously described.

Judges

The five judges were made up of two experienced orthodontists, one orthodontic resident, one freshman dental student, and one expert artist who specialized in cephalometrics. The last judge, Mr William Golden, has been dead for a number of years, but his expert tracings of the sample lateral cephalometric radiographs were available for our study. One investigator digitized all the landmarks.

Method

Each lateral cephalometric radiograph was traced twice by five judges and four fiducial reference marks, and twenty one anatomical reference points were identified. The selected points were clearly marked as a fine pencil dot; or an "X". In areas of the tracings where confusion my have arisen as to the identity of multiple points close to one another, these ambiguous points where clearly labeled. To eliminate memory bias of the judges, the repeated tracing occurred at least two weeks after the initial tracing. The points used are listed below

A. Fiducial Reference Marks

1. **Fiducial 1**

   This is the pin hole in the radiograph located at the right superior corner of the film.

2. **Fiducial 2**

   This is the pin hole in the radiograph located at the right inferior corner of the film

3. **Fiducial 3**

   This is the pin hole in the radiograph located at the left inferior corner of the film

4. **Fiducial 4**

   This is the pin hole in the radiograph located at the left superior corner of the film.

B. Anatomic Landmarks

1. **Sella**

   Sella Turcica, the midpoint of the pituitary fossa as determined by inspection.

2. **N**

   Nasion, the most antero-inferior point on the frontal bone at the nasofrontal suture.

3. **Or**

   Orbitale, the most inferior point of the orbital margin, using the more prominent of the orbital images.

4. **ANS**

   Anterior Nasal Spine, the anterior tip of the sharp bony process of the maxilla at the lower margin of the anterior nasal opening.

5. **PNS**

   Posterior Nasal Spine, the posterior spine of the palatine bone containing the hard palate.

6. **UI edge**

   Upper central incisor edge, the tip of the incisal edge of the more anteriorly placed upper central incisor.

7. **UI apex**

   Upper central incisor apex, the point of intersection between the long axis of the more anteriorly positioned upper central incisor and the contour of the tooth's root and curvature

8. **LI edge**

   Lower central incisor edge, the tip of the incisal edge of the more anteriorly placed lower central incisor.

9. **LI apex**

   Lower central incisor apex, the point of intersection between the long axis of the most anteriorly positioned lower incisor and the contour of the tooth's root end curvature.

10. **UFM cusp**

Upper first molar mesiobuccal cusp, (the average value between the left and right), the most occlusal point of the mesiobuccal cusp of the upper first molar.

11. **UFM apex**

Upper first molar mesiobuccal root apex, (the average value between the left and the right), the most apical point on the mesiobuccal root of the upper first molar.

12. **LM cusp**

Lower first molar mesiobuccal cusp (the average value between the left and the right) of the most occlusal point mesiobuccal cusp of the lower first molars. Note the more mesially positioned will usually be the right, and the more distally positioned will usually be the left.

13. **LM apex**

Lower first molar mesiobuccal root apex, (the average value between the left and the right), the most apical point of the mesiobuccal root of the lower first molar.

14. **A**

Point A, the deepest point on the curvature of the surface of the surface of the maxillary bone between ANS (Anterior Nasal Spine) and the alveolar crest of the upper central incisor

15. **B**

Point B, the deepest point on the curvature of the anterior border of the mandible between Pogonion and the alveolar crest of the lower central incisor.

16. **Pg**

Pogonion, the most anterior point on the bony chin at the midline

17. **Upper Go**

Upper Gonion, or Gonion for the left side This is a point on the curvature of the angle of the mandible located by bisecting the angle by lines tangent to the posterior ramus and the inferior border of the mandible The lowest point on the curvature of the angle of the mandible where the body of the mandible meets the ramus. Estimates were made for both sides of the mandible The more superior value is considered to represent the left side.

18. **Lower Go**

Lower Gonion, or Gonion for the right side. The lowest point on the curvature of the angle of the mandible where the body of the mandible meets the ramus. Estimates were made for both sides of the mandible. The more inferior value is considered to represent the right side

19. Mechanical porion

Found by using Broadbent template. Place template over ear rods, and head support, then mark the point representing mechanical Porion

20. Condyle

Condyle point, the perceived central point of the mandibular condyle, averaged between left and right sides.

21. Me

Menton, the most inferior point on the mandible at the symphysis

The actual instructions given to the judges are given in Appendix A These landmarks were chosen as they had previously been selected as a good representation of all points used in cephalometrics (Baumrind et al 1971) A customized computer program was used to convert the raw data from the tracings into the x,y coordinate form. The measurement of each data set was completed twice, to establish measurement error. Each radiographic tracing had effectively four sets of measurements associated with it The principal investigator performed all of the digitizing on a digitizing tablet (Summagraphics ® Summasketch ™). Having one operator perform all of the digitizing reduced random error from points being digitized in a incorrect sequence, and reduced errors associated with incorrect use of the software measurement program.

The digitizing tablet has an inherent accuracy of 0 001 mm A puck with a cross hair is placed over each point on the tracing then the x,y coordinates of the point are computed. To assist in recognition of the points, a piece of clean white paper was first taped to the tablet, so when the tracing was taped over this, the marked points were more easily recognized This sheet of paper also assisted in marking the actual digitizing area of the tablet. Each point was digitized with the operator directly over the point. All cephalometric points were digitized in the same order. Five radiographs were measured using a calibrated ruler to verify the data generated from the computer.

The digitizing tablet and software measurements proved to be at least as accurate as the measurements taken with a calibrated ruler

The software program (Plotodont) used two of the fiducial data points as reference markers. The lower left fiducial marker was defined as the origin of the x,y coordinate axis system, and assigned the coordinates 0,0. A line was drawn to the upper left hand fiducial reference mark, and the x coordinate of this line was defined as zero. This procedure defined the y axis, the origin being at the lower left fiducial reference mark, and continuing through the upper left hand fiducial reference mark, defined as zero on the x axis The x axis was defined as a perpendicular line originating from the lower left hand fiducial reference mark. After the axes were defined, each landmark was assigned an x,y coordinate. Because the fiducial reference marks were always the same, landmarks used in this study were referenced to the same axes. Landmark locations for each film were able to be compared between the same judge at different timepoints, and between different judges

Following the double determination digitization of each tracing, the principal investigator evaluated both sets of measurements for any gross discrepancies, and if measurement error was greater than 0.3 mm, the readings were disregarded, and both measurements were repeated. Measurement error was consequently below the 0.3 mm maximum This maximum error value, 0.3 mm, was chosen as this is the thickness of the pencil lead used in the tracing process. Once two measurements of an individual tracing were determined to be within the measurement error range, a printed copy was made of the coordinates of all the points, and a ASCII text file saved with the same results. A hard copy of the results was made as a backup copy, in case the electronic file became corrupted

Once all the data from the tracings had been digitized, the ASCII text files containing this data were imported into a spreadsheet program (Microsoft Excel 4 0) Each individual tracing had been measured twice, and these measurements were within a maximum error range of 0.3 mm. These measurement values were averaged to obtain mean measured values for each tracing. The upper and lower gonial angles were averaged to produce a single result for the gonial angle after Baumrind (1971). For each radiograph, the landmark locations for each point were averaged from a summation of each investigators double determination tracings, and an average x,y coordinate for each landmark was located within an individual film. The deviation for each point in an x and y direction from the mean was calculated, and these results were then plotted. Using this procedure it was possible to compare a particular point location from each of the twenty five radiographs, and plot all the deviations from their individual means to obtain an overall picture of the spread of a particular landmark when it is located by different operators, and by the same operator at different times

Objective testing of Spatial Resolution from the Monitor

The result of the subjective test of spatial resolution using the two matched monitors indicated that the medium resolution image (768 pixels X 512 pixels) offered more information to the clinician than the low resolution image (384 pixels X 256 pixels). This test also showed that the majority of judges could see no difference between the mid and high resolution images. Following the determination of the optimal spatial resolution from the initial subjective testing of the computer generated images, the spatial resolution of 768 pixels X 512 pixels was tested objectively to determine if this given resolution was adequate to accurately identify cephalometric landmarks.

Sample

The sample of 25 radiographs used previously were displayed on a computer monitor using a commercially available graphics program (Adobe Photoshop Version 2.1., Adobe Systems Inc., Mountain View, California). All of the images were displayed at a spatial resolution of 768 pixels X 512, and viewed on the monitor at a ratio of 2·1. This was the same viewing resolution used in the subjective monitor test Some panning across the screen was necessary to view the entire image of the lateral cephalometric radiograph.

Using the line tool available with Adobe Photoshop V 2 1, the cursor was able to be anchored at the lower left fiducial point. The cursor cross hair was then placed over the 21 anatomic landmarks, and the three remaining fiducial reference marks. Using the information window in Adobe Photoshop V 2 1, it was possible to record to two decimal places a delta x, and delta y value of all of the landmarks in relation to the lower left hand fiducial of the radiograph image. Each measurement value was manually recorded on a sheet, as it was not possible to electronically compile these measurements with this software program

Two judges from the group of the initial four judges who objectively judged the original radiographs participated in this test Double determinations were performed at least two weeks apart to minimize memory bias of the judges. When a landmark was located the measurement was immediately visible on the screen, so digitizing error was not a consideration in this test. The test exclusively dealt with landmark location

The data was directly entered into a spreadsheet program (Microsoft Excel 4 0) for data organization. The screen measurements were not in scale with the actual measurements from the radiographs. The actual distances between the fiducial

reference points was known, and a conversion factor was established in the x direction, and in the y direction for each film All of the measurements were transformed by these conversion factors to enable each x,y coordinate to be in a true millimeter scale, and comparable between data sets.

The same procedure was used on this data as in the data from the original radiographs For each radiographic image, a given point was averaged, and all the individual measurements for this point were subtracted from the mean to give each point a value of its deviation from the mean The spread of values for each point was then able to plotted together with the same points from each of the 25 films to view a cumulative landmark location error for each specific point tested. These graphs were then compared to the landmark location error from the original radiographs. Scattergram graphs were used to graphically depict the true error involved in landmark location both in the horizontal (x axis) and vertical (y axis) planes.

RESULTS

Spatial resolution of cephalometric radiographs was examined by a subjective test to determine the level of spatial resolution a clinician finds offers the most information from a digital representation of a lateral cephalometric radiograph. The digital representations of 25 lateral cephalometric radiographs were then tested at this level of spatial resolution, to objectively to identify 21 anatomic and four fiducial landmarks. The landmarks were located in a x,y coordinate fashion, and then compared to the same landmarks identified from the original silver halide radiographic films. The data were analyzed according to individual reference points located, and graphically depicted to describe the error involved in landmark identification

Spatial Resolution Subjective Test - Monitor Bias

The results of the subjective spatial resolution test are presented in Table 1 and shown graphically in figure 1 Each judge was given ten identical image pairs to examine at the same resolution. Each judge was able to make ten choices as to whether identical images were biased by either Monitor A or Monitor B. Examining the aggregate results of all the judges (a total of 280 responses). 55.3% selected no difference between the images, 15% selected the image on Monitor A as offering more information; and 29.64% selected the image on Monitor B as offering more information Monitor bias was not significant in determining image selection by the judges.

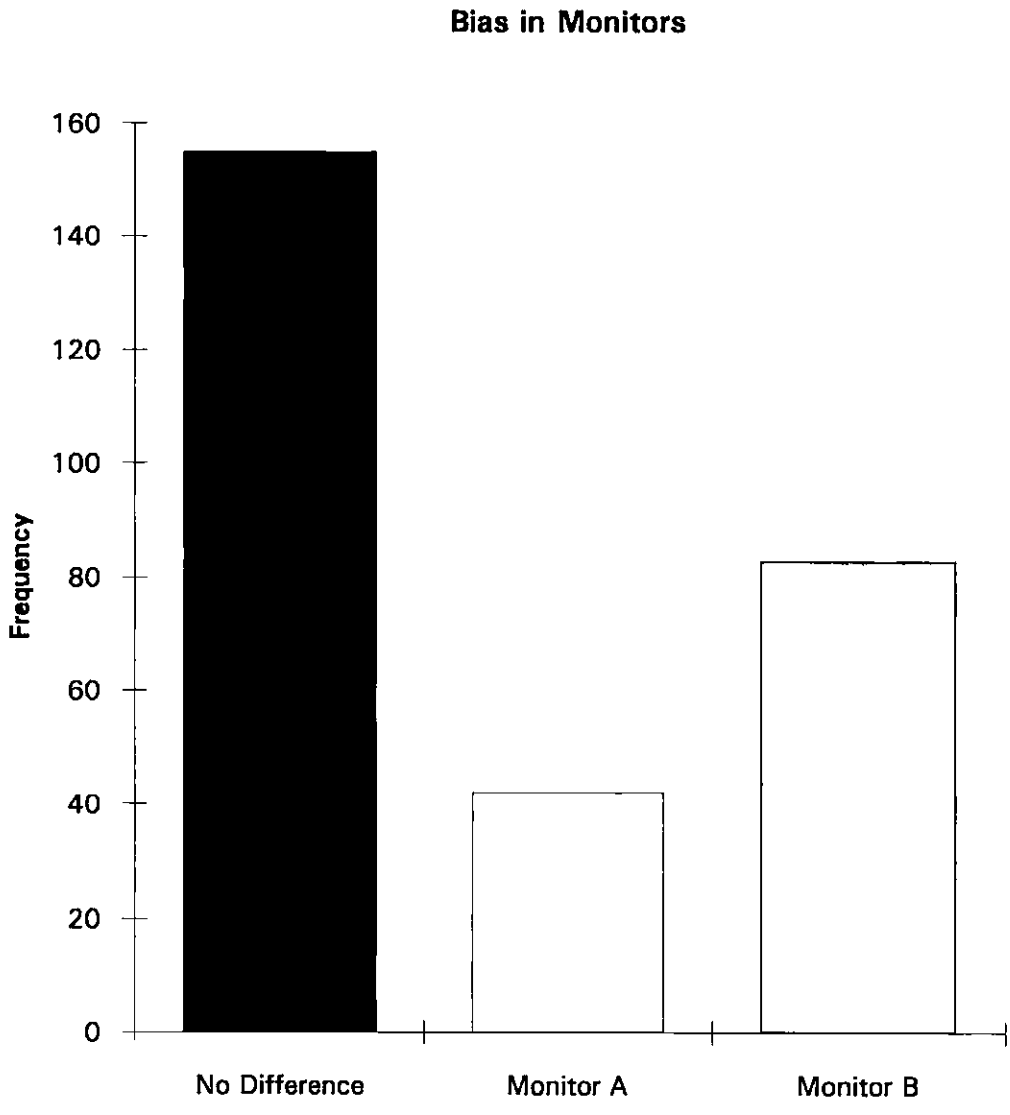


Figure 1. Comparison of identical images shown on both monitors to determine monitor bias See Table 1 for numerical data.

Table 1.--Monitor Bias as a frequency distribution

Judges	Category	No Difference	Monitor A	Monitor B
1	Resident Orthodontist	7	2	1
2	Resident Oral Surgeon	6	1	3
3	Orthondontist	0	8	2
4	Resident Orthodontist	8	0	2
5	Orthodontist	9	1	0
6	Orthodontist	6	1	3
7	Resident Orthodontist	10	0	0
8	Resident Orthodontist	7	1	2
9	Resident Orthodontist	5	0	5
10	Resident Orthodontist	1	7	2
11	Resident Orthodontist	7	0	3
12	Resident Orthodontist	3	0	7
13	Resident Orthodontist	2	5	3
14	Resident Orthodontist	1	4	5
15	Resident Orthodontist	7	0	3
16	Orthodontist	5	0	5
17	Orthodontist	9	1	0
18	Orthodontist	1	1	8
19	Orthodontist	6	0	4
20	Orthodontist	4	0	6
21	Orthodontist	2	0	8
22	Orthodontist	4	4	2
23	Orthodontist	8	0	2
24	Oral Surgeon	9	1	0
25	Orthodontist	8	0	2
26	Orthodontist	7	0	3
27	Orthodontist	4	4	2
28	Oral Surgeon	9	1	0
Total		**155**	**42**	**83**
Frequency		**55.36%**	**15.00%**	**29.64%**

The same radiograph displayed on both monitors to assess any bias present in the computer monitors.

Low Versus Medium Resolution

This test examined the ability of the judges to choose which radiograph offered the most information when presented with the same lateral cephalometric radiograph, with one image viewed at a low resolution (384 pixels X 256 pixels) on Monitor A, and a medium resolution (768 pixels X 512 pixels) image displayed on Monitor B. Each judge was given five different radiographic pairs to view in this fashion. The aggregate results of all the judges (a total of 140 responses) are listed in table 2 and figure 2. The vast majority of judges (91.43%) found the medium resolution (512 pixels X 768 pixels) image offered more information than the low resolution (256 pixels X 384 pixels) image. Only 1 43% of responses found there to be no difference in the information available from the low versus medium resolution image 7.14% of the judges found the low resolution image to offer more information than the medium resolution image. This low percentage figure corresponds to 10 decisions being made out of 140. In this test, each judge was shown five different radiographic pairs This meant an individual judge was able to make a maximum of five choices Seven out of the ten decisions judging the low resolution film came from two judges

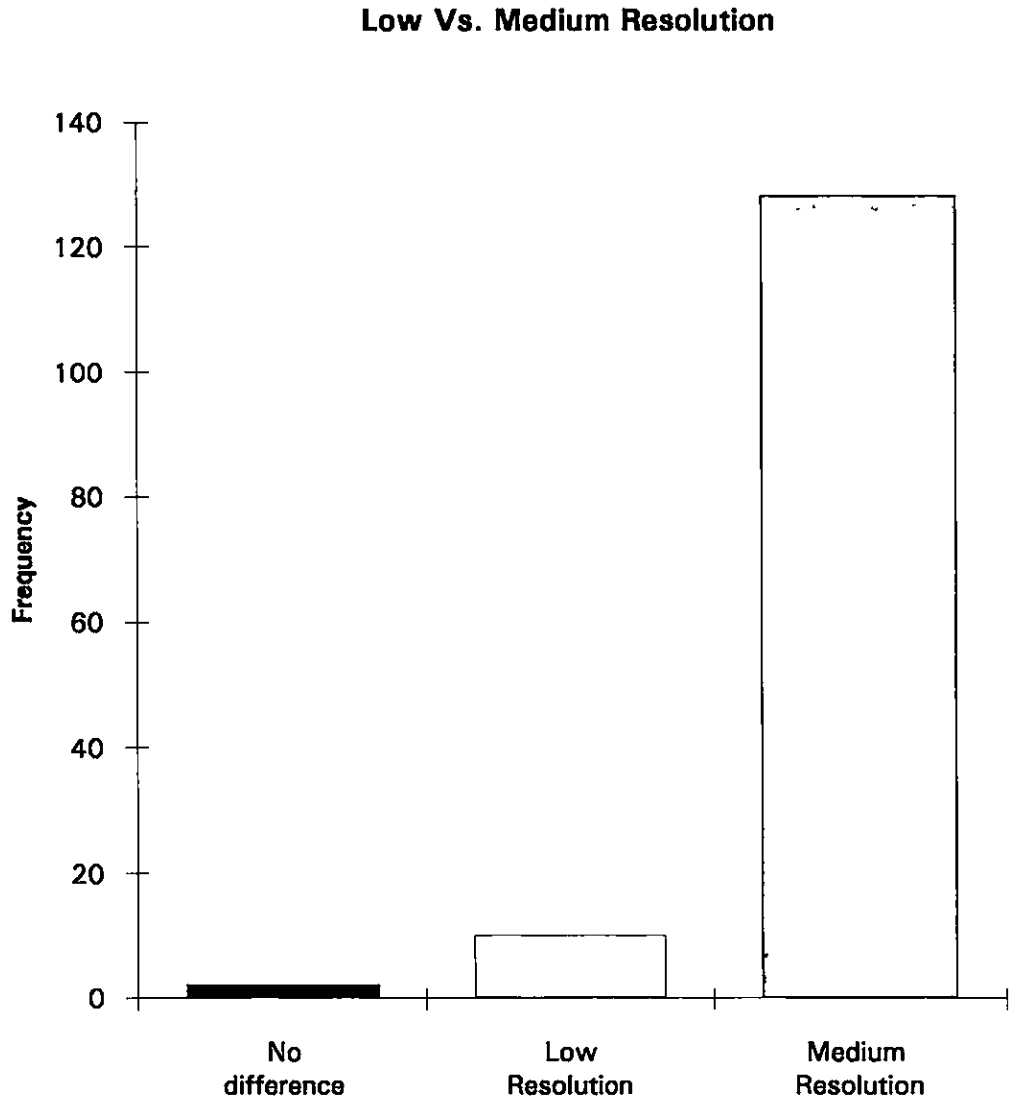


Figure 2. Comparison of the low resolution and the medium resolution cephalometric image to determine which image offers the most information. See Table 2 for numerical data.

Table 2.-- Low Versus Medium Resolution Cephalometric Radiographic Images

Judges	Category	No difference	Low Resolution	Medium Resolution
		1	0	4
1	Orthodontic Resident	0	0	5
2	Oral Surgery Resident	0	4	1
3	Orthondontist	0	0	5
4	Orthodontic Resident	0	0	5
5	Orthodontist	0	0	5
6	Orthodontist	0	0	5
7	Orthodontic Resident	0	0	5
8	Orthodontic Resident	1	0	4
9	Orthodontic Resident	0	1	4
10	Orthodontic Resident	0	0	5
11	Orthodontic Resident	0	0	5
12	Orthodontic Resident	0	3	2
13	Orthodontic Resident	0	0	5
14	Orthodontic Resident	0	0	5
15	Orthodontic Resident	0	0	5
16	Orthodontist	0	0	5
17	Orthodontist	0	2	3
18	Orthodontist	0	0	5
19	Orthodontist	0	0	5
20	Orthodontist	0	0	5
21	Orthodontist	0	0	5
22	Orthodontist	0	0	5
23	Orthodontist	0	0	5
24	Oral Surgeon	0	0	5
25	Orthodontist	0	0	5
26	Orthodontist	0	0	5
27	Orthodontist	0	0	5
28	Oral Surgeon	0	0	5
Total		2	10	128
Frequency		1.43%	7.14%	91.43%

Five different radiographs with each radiograph compared at a low (384 pixels X 256 pixels) resolution and a medium (768 pixels X 512 pixels) resolution

Medium Versus High Resolution

This test examined the ability of the judges to choose which radiograph offered the most information when presented with the same lateral cephalometric radiograph viewed at a medium resolution (512 pixels X 768 pixels) on one monitor, and at a high resolution (1024 pixels X 1536 pixels) on the other monitor. There was a total of 280 decisions in this test. Almost half the judges, 47.14%, found there to be no information difference between the medium and high resolution. 38.21% of the judges found the medium resolution image to offer more information than the higher resolution image. The remaining 14.64% of the judges found the higher resolution image to offer more information than the lower resolution film. These data are presented in Table 3 and graphically depicted in figure 3

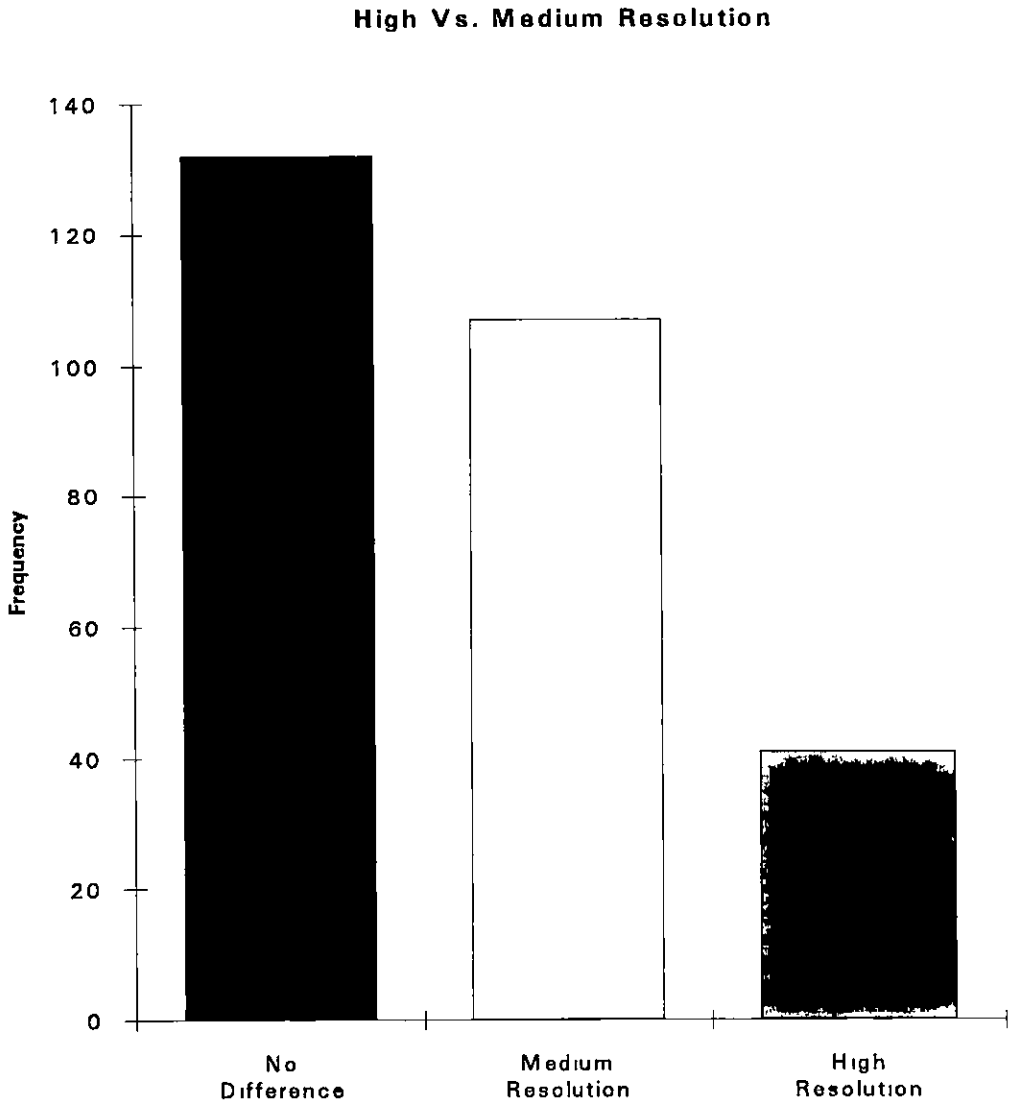


Figure 3. Comparison of the medium resolution and the high resolution cephalometric image to determine which image offers the most information See Table 3 for numerical data

Table 3.-- High Versus Medium Resolution Cephalometric Radiographic Images

Judges	Category	No Difference	Medium Resolution	High Resolution
1	Orthodontic Resident	5	5	0
2	Oral Surgery Resident	5	5	0
3	Orthondontist	0	8	2
4	Orthodontic Resident	7	1	2
5	Orthodontist	9	1	0
6	Orthodontist	3	7	0
7	Orthodontic Resident	9	1	0
8	Orthodontic Resident	2	5	3
9	Orthodontic Resident	4	3	3
10	Orthodontic Resident	2	5	3
11	Orthodontic Resident	4	4	2
12	Orthodontic Resident	3	5	2
13	Orthodontic Resident	4	3	3
14	Orthodontic Resident	2	8	0
15	Orthodontic Resident	4	6	0
16	Orthodontist	5	5	0
17	Orthodontist	9	1	0
18	Orthodontist	2	5	3
19	Orthodontist	2	6	2
20	Orthodontist	1	6	3
21	Orthodontist	4	3	3
22	Orthodontist	5	2	3
23	Orthodontist	7	1	2
24	Oral Surgeon	9	1	0
25	Orthodontist	7	1	2
26	Orthodontist	4	6	0
27	Orthodontist	5	2	3
28	Oral Surgeon	9	1	0
Total Frequency		132 47.14%	107 38.21%	41 14.64%

Five different radiographs with each radiograph compared at a high (1536 pixels X 1024 pixels) resolution and a medium (768 pixels X 512 pixels) resolution

Spatial Resolution Objective Test

The x,y coordinates were computed for the twenty one anatomic and four fiducial landmarks located on the twenty five original radiographs. Five judges performed this test.

Judge 1. Orthodontic resident,
Judge 2: Junior Dental Student;
Judge 3 Experienced Orthodontist;
Judge 4· Experienced Orthodontist;
Judge 5. Cephalometric Expert.

All the judges, except Judge 5, located these landmarks on two separate occasions To enable the extensive data of x,y coordinates to be interpreted, the x and y coordinates from all the judges observations for each individual point (e.g. Sella) for each radiograph were averaged. Each judges observation was then plotted as a deviation from the mean, both in the x and in the y axes for each radiograph Consequently, a plot was made with the origin being the mean value off all the judges deviations from the mean for a individual point, that is 0,0. This procedure graphically depicted the error in landmark location for an individual point for a particular radiograph. Because the individual judges observations had been converted to values representing deviations from the mean, it was possible to combine all the data for an individual point from all twenty five radiographs. This procedure graphically represented the error in landmark location for an individual point from all 25 radiographs Graphs are presented for each of the 21 anatomic landmarks, and for 3 of the fiducial landmarks. Fiducial 4 was not graphed as this point is by definition the origin of all the measurements (0,0).

Two judges (Judge 1 and Judge 4) were involved in selecting landmarks from the computer monitor. The x,y coordinate landmark locations were plotted as

deviations from the mean using the same method previously described for the original radiographs. These data were graphically depicted to illustrate the error in landmark location using a computer monitor at the spatial resolution of 768 pixels X 512 pixels. To illustrate the differences in landmark location error from the computer monitor and the original radiographs, each plot of the monitor data is accompanied by Judge 1 and Judge 4 original radiographic data.

Intra Judge Reliability

To assess intra judge reliability, the data for each individual judge for a particular point was compared from time point 1 to time point 2. Pearson's Correlation Coefficient and the Students t test for paired samples are reported for each point. For the monitor tests of judge 1 and 4, these judges were coded 1m and 4m respectively to differentiate the monitor from the original radiograph test

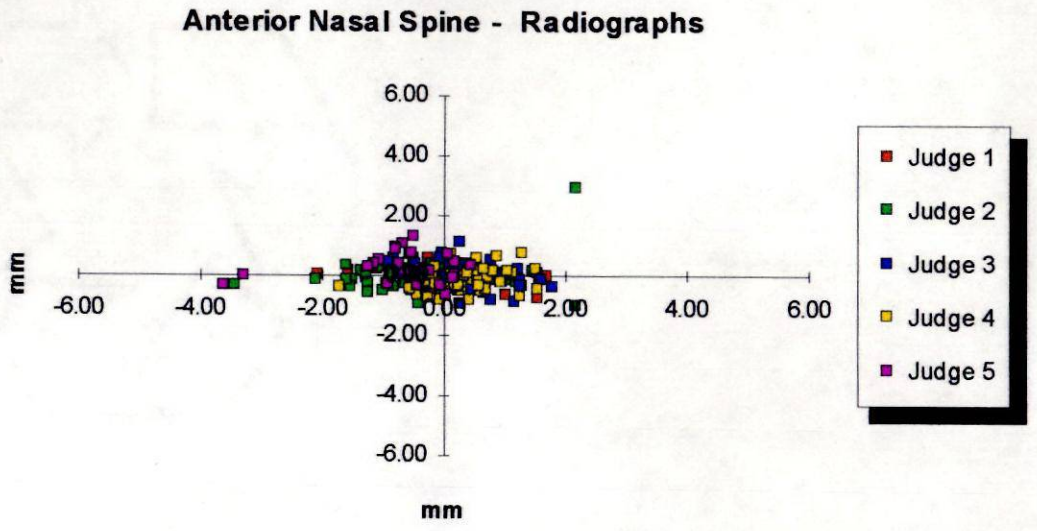


Figure 4. Error associated with the location of Anterior Nasal Spine from all twenty five radiographs.

Table 4.-- Intra-Examiner Reliability Anterior Nasal Spine

Anterior Nasal Spine x coordinate. Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.995	0.990	79.83	0.42	0.6790
2	0.987	0.975	76.69	-0.16	0.8723
3	0.995	0.989	79.24	0.69	0.4945
4	0.997	0.995	81.23	2.48	0.0205
1m	0.998	0.997	89.92	0.94	0.3564
4m	0.998	0.996	89.79	0.63	0.5344
Anterior Nasal Spine y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.999	0.998	107.39	-0.76	0.4561
2	0.997	0.994	106.86	1.50	0.1469
3	0.998	0.996	108.39	1.09	0.2862
4	0.999	0.997	106.56	0.94	0.3573
1m	0.999	0.999	110.77	-0.43	0.6685
4m	0.999	0.997	109.66	-0.41	0.6882

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

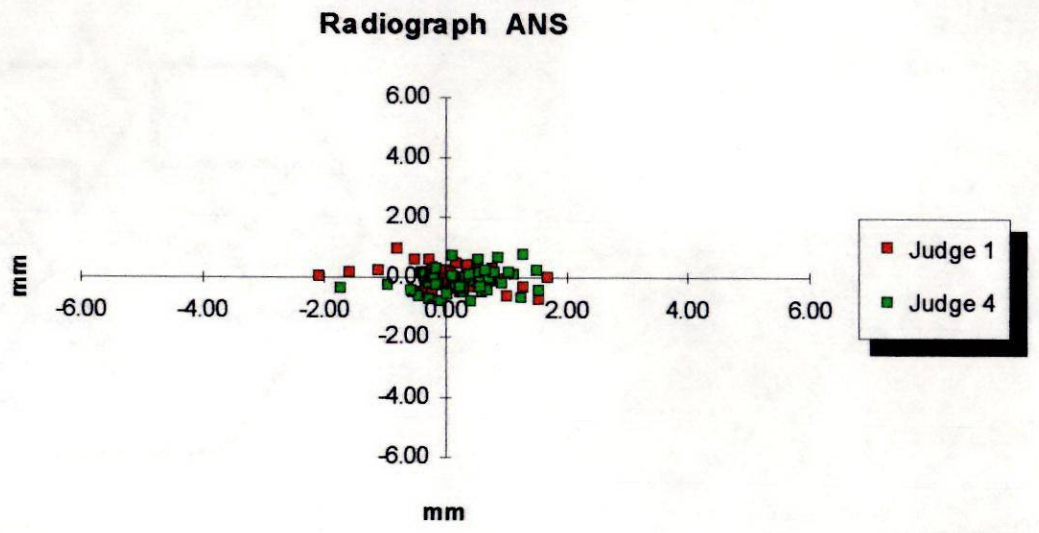


Figure 5. Error associated with the location of ANS for Judge 1 and 4 from the radiographs.

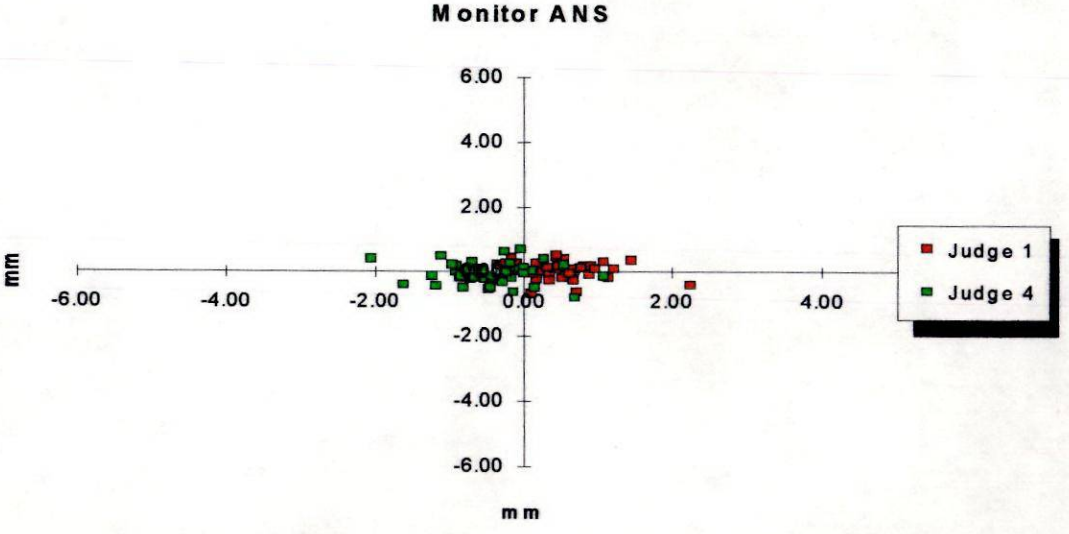


Figure 6. Error associated with the location of ANS for Judge 1 and 4 from the computer monitor.

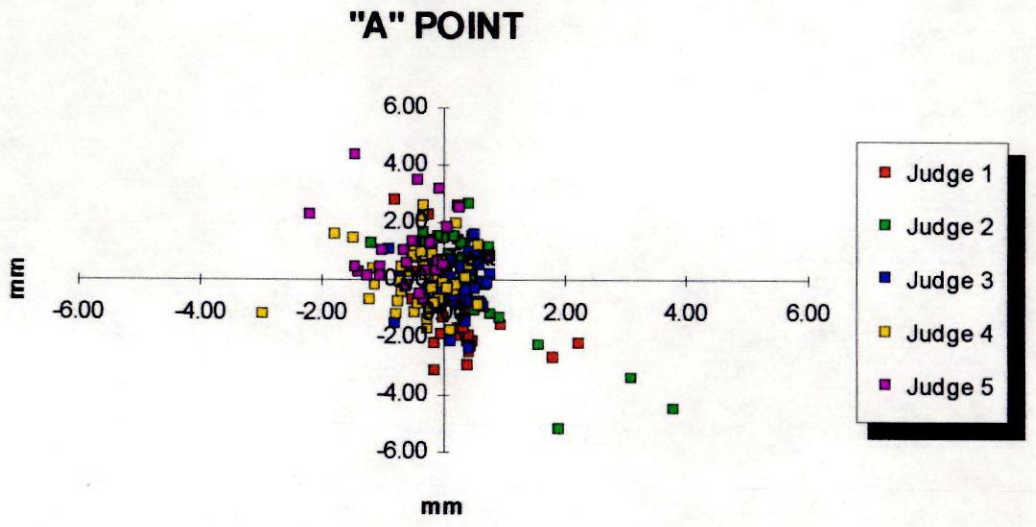


Figure 7. Error associated with the location of 'A' Point from all twenty five radiographs

Table 5.-- Intra-Examiner Reliability 'A' Point

"A" Point x coordinate Alpha= 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	1.00	0.99	80.86	-0.85	0.4035
2	0.99	0.99	83.64	-2.23	0.0355
3	1.00	1.00	78.60	0.94	0.3583
4	1.00	0.99	80.86	-0.85	0.4035
1m	1.00	1.00	88.44	0.00	1.0000
4m	1.00	1.00	83.76	1.39	0.1773
"A" Point y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.99	0.97	93.05	0.46	0.6503
2	0.98	0.97	101.64	2.32	0.0293
3	1.00	0.99	97.56	1.07	0.2976
4	0.99	0.97	95.57	0.09	0.9294
1m	1.00	0.99	95.79	0.19	0.8487
4m	1.00	0.99	96.24	-0.85	0.4045

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

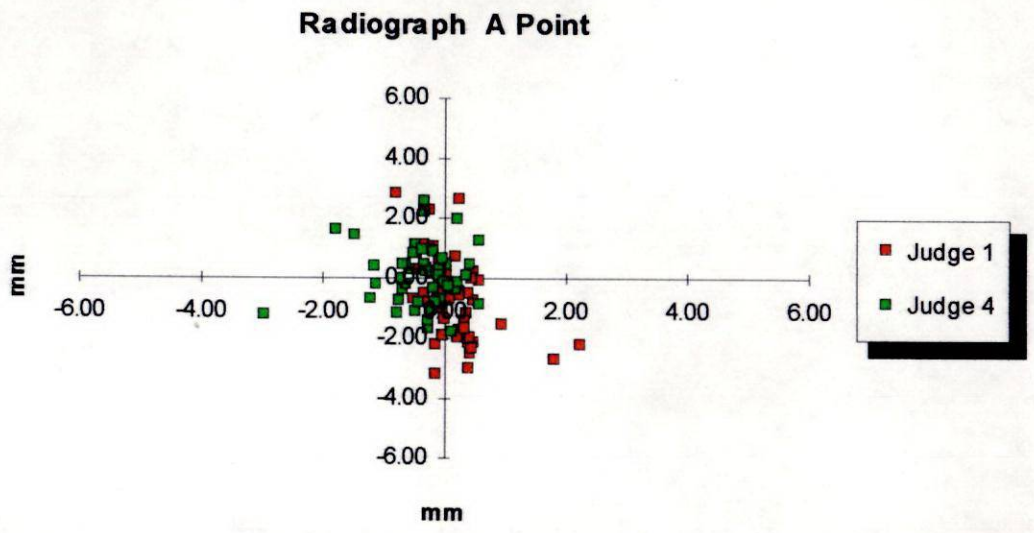


Figure 8. Error associated with the location of 'A' Point for Judge 1 and 4 from the radiographs.

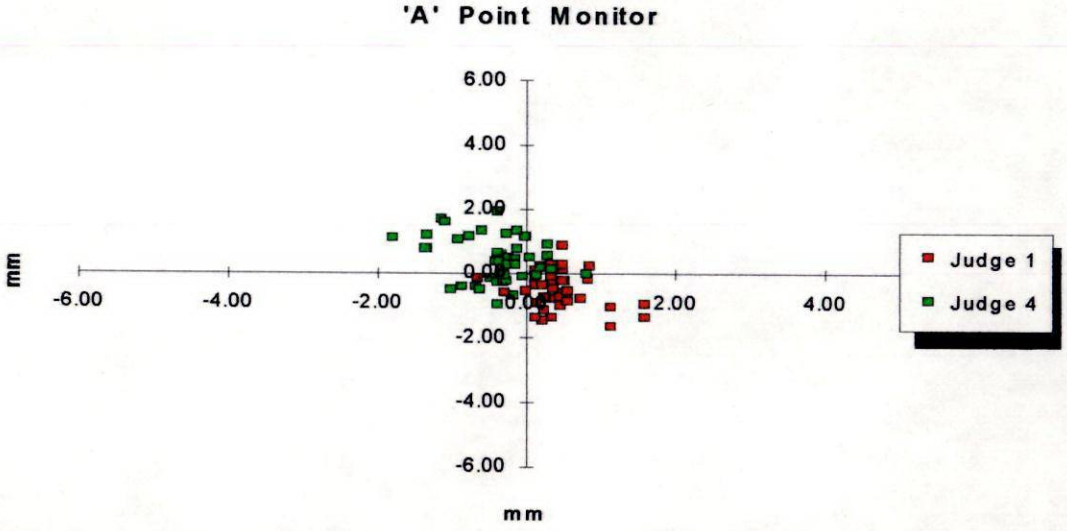


Figure 9. Error associated with the location of 'A' Point from Judge 1 and 4 for the computer monitor.

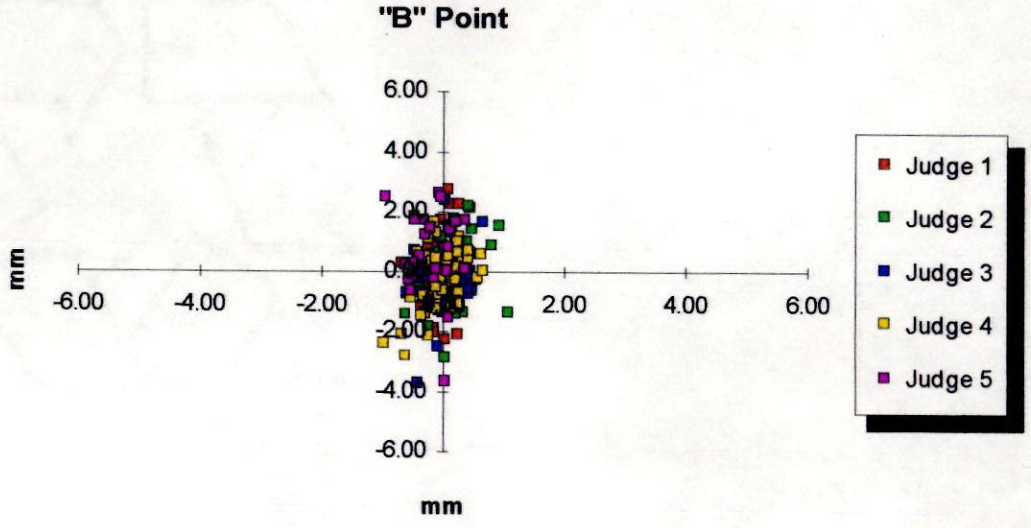


Figure 10. Error associated with the location of 'B' Point from all twenty five radiographs.

Table 6.-- Intra-Examiner Reliability 'B' Point

"B" Point x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.999	0.999	88.51	-1.28	0.2132
2	0.999	0.997	87.10	-2.59	0.0162
3	0.999	0.998	105.49	0.07	0.9477
4	0.999	0.997	86.83	0.79	0.4360
1m	1.000	1.000	93.24	-0.65	0.5197
4m	0.999	0.998	94.49	0.71	0.4831
"B" Point y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.981	0.963	60.63	1.19	0.2474
2	0.984	0.968	77.78	-0.81	0.4237
3	0.995	0.990	102.89	-1.90	0.0739
4	0.985	0.971	79.93	0.12	0.9048
1m	0.986	0.973	91.79	-1.57	0.1301
4m	0.996	0.993	87.30	3.94	0.0007

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

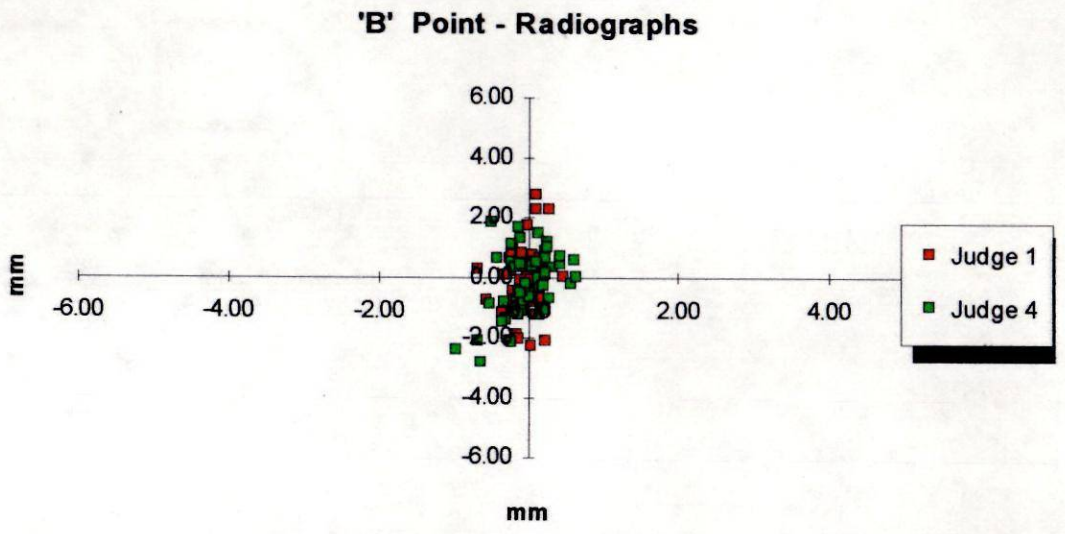


Figure 11. Error associated with the location of 'B' Point for Judge 1 and 4 from the radiographs.

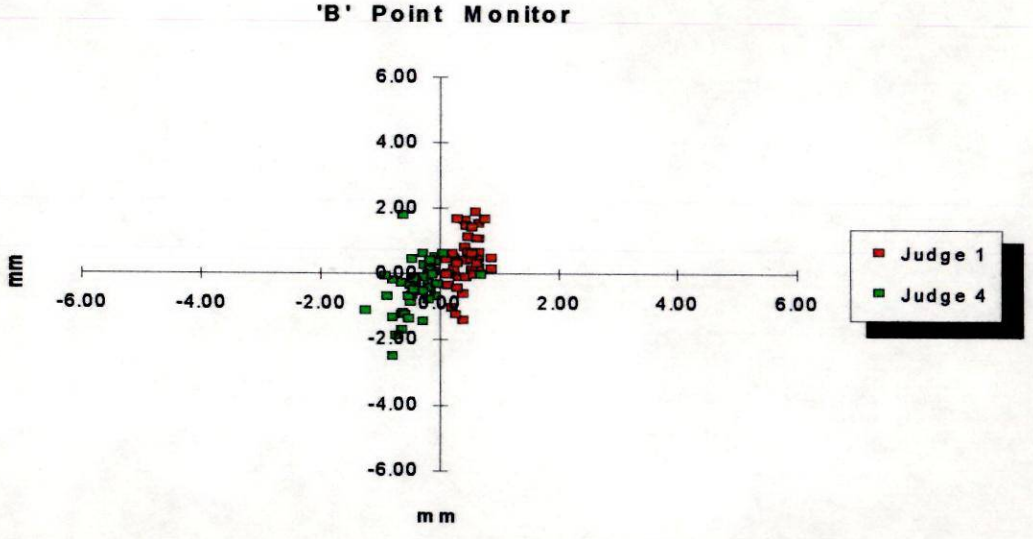


Figure 12. Error associated with the location of 'B' Point for Judge 1 and 4 from the computer monitor.

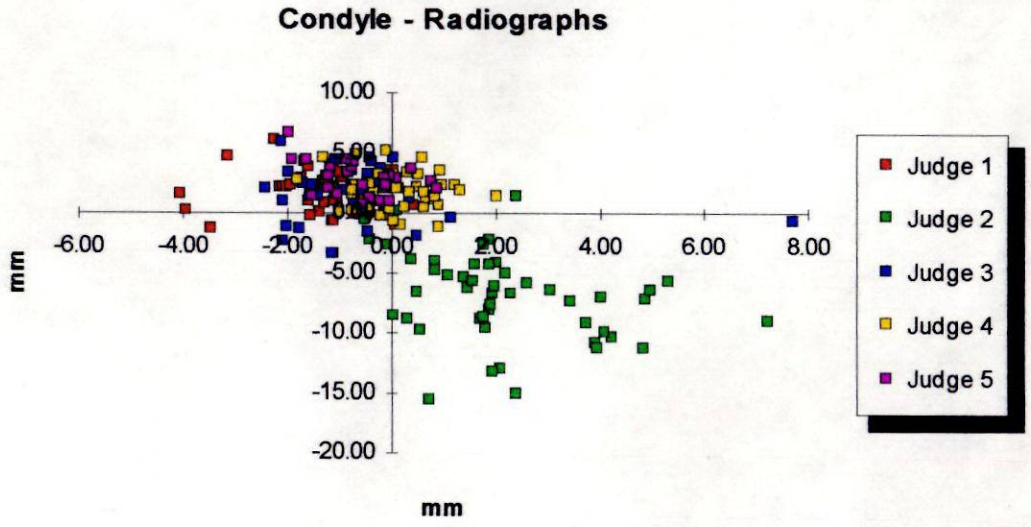


Figure 13. Error associated with the location of Condylion from all twenty five radiographs.

Table 7.-- Intra-Examiner Reliability Condylion

Condyle x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.998	0.996	66.71	-0.11	0.9129
2	0.971	0.942	66.90	2.55	0.0178
3	0.418	0.175	109.90	1.51	0.1558
4	0.996	0.993	67.86	-1.51	0.1429
1m	0.999	0.998	75.22	-0.84	0.4639
4m	0.995	0.990	64.92	1.00	0.3277
Condyle y Coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.984	0.969	119.22	-0.08	0.9358
2	0.906	0.822	119.93	-3.23	0.0036
3	0.082	0.007	22.36	1.62	0.1284
4	0.996	0.991	141.73	0.03	0.9757
1m	0.999	0.999	185.77	-0.82	0.4726
4m	0.990	0.981	127.86	-0.44	0.6642

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

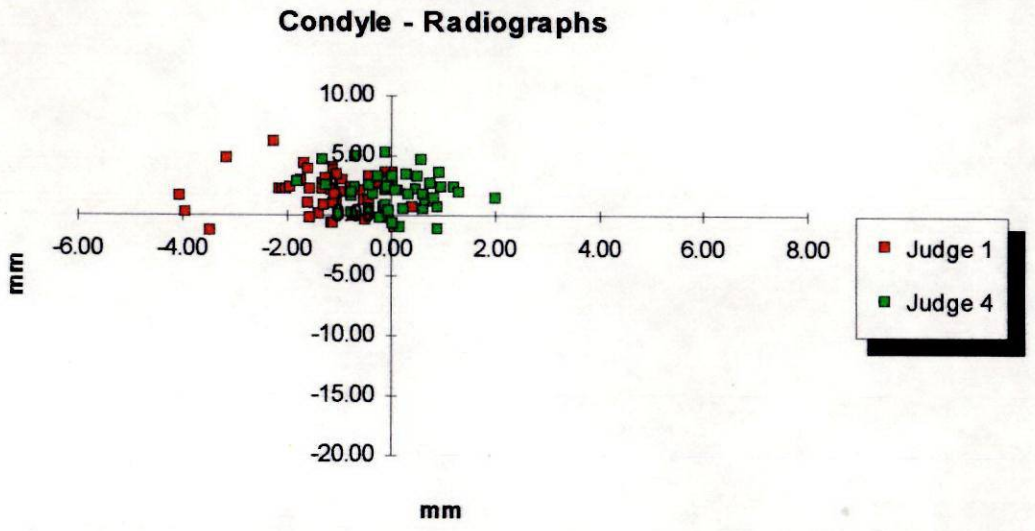


Figure 14. Error associated with the location of Condylion for Judge 1 and 4 from the radiographs.

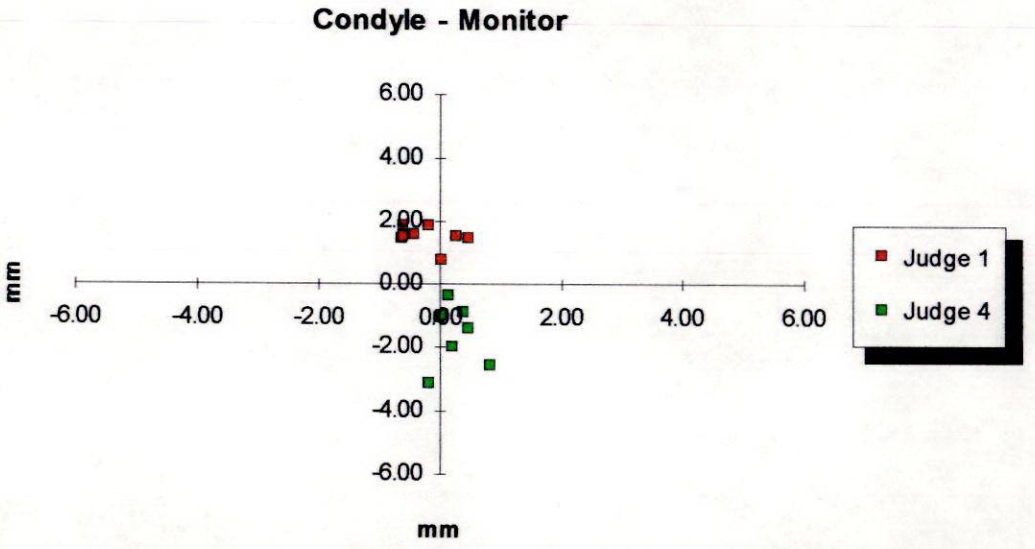


Figure 15. Error associated with the location of Condylion for Judge 1 and 4 from the computer monitor.

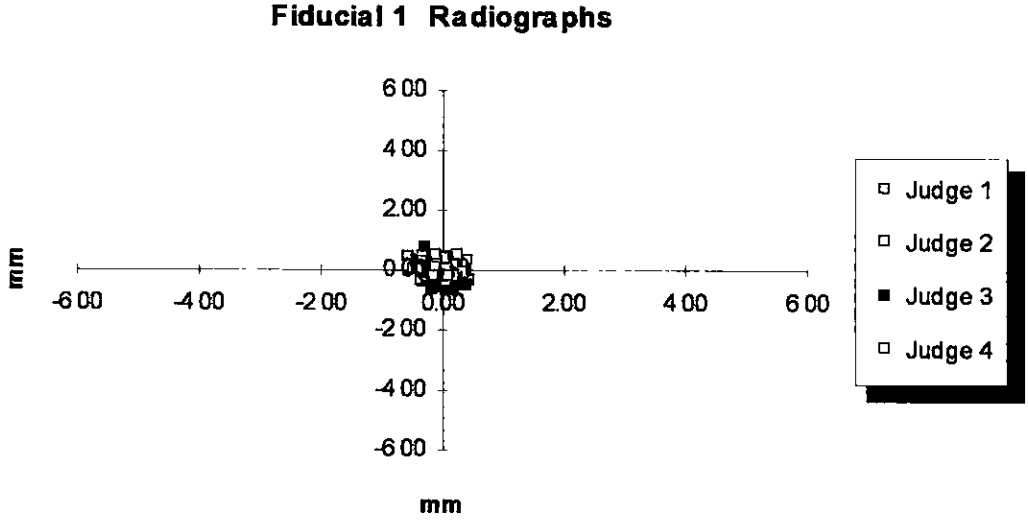


Figure 16. Error associated with the location of Fiducial 1 from all twenty five radiographs.

Table 8.-- Intra-Examiner Reliability Fiducial 1

Fiducial 1 x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.441	0.194	0.01	0.07	0.9456
2	-0.141	0.020	-0.01	-2.32	0.0289
3	-0.087	0.008	0.00	-1.03	0.3155
4	-0.323	0.104	-0.02	-0.74	0.4645
1m	0.969	0.939	1.00	1.56	0.1336
4m	-0.323	0.104	-0.02	-0.74	0.4645
Fiducial 1 y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.199	0.040	0.01	2.30	0.0302
2	0.405	0.164	0.04	-0.47	0.6435
3	-0.155	0.024	-0.01	-2.87	0.0086
4	0.363	0.132	0.02	-2.52	0.0187
1m	0.958	0.918	1.57	0.27	0.7863
4m	0.363	0.132	0.02	-2.52	0.0187

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

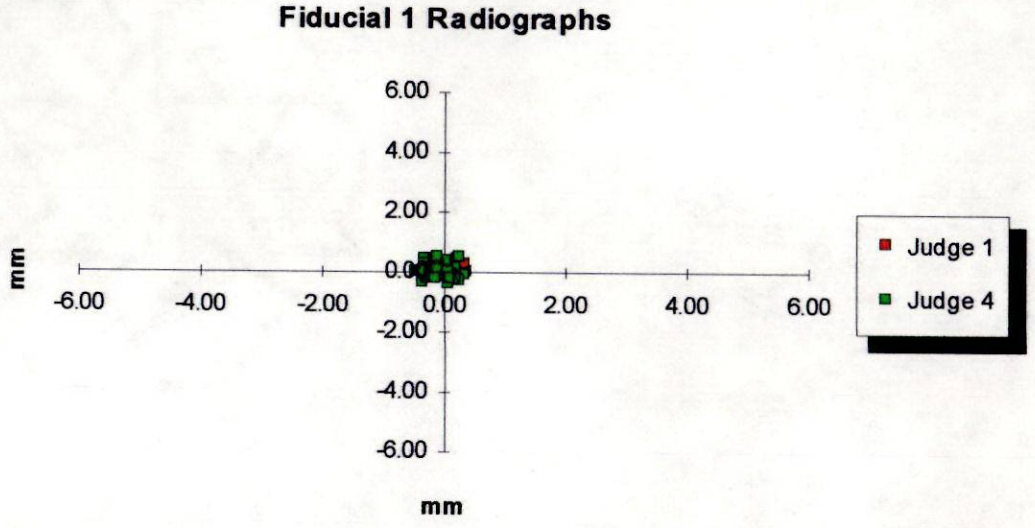


Figure 17. Error associated with the location of Fiducial 1 for Judge 1 and 4 from the radiographs.

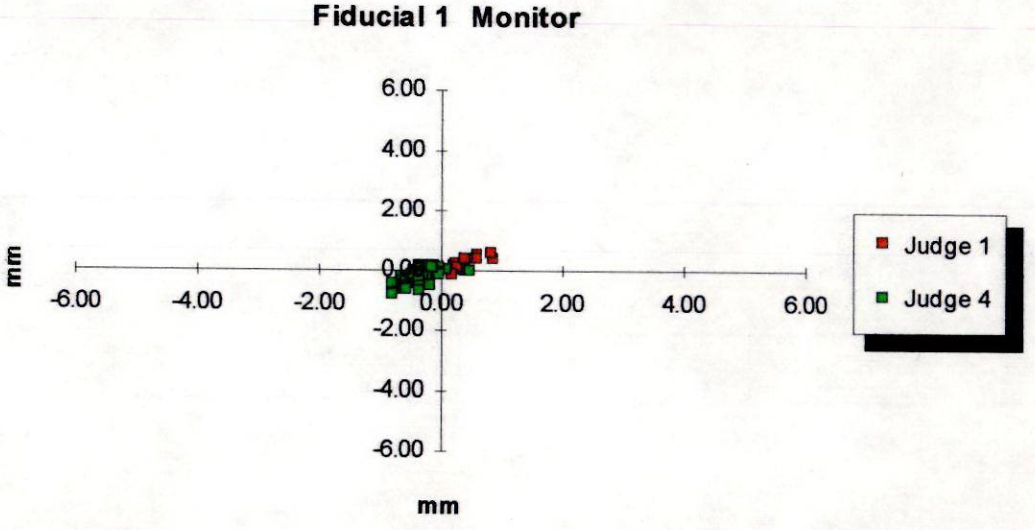


Figure 18. Error associated with the location of Fiducial 1 for Judge 1 and 4 from the computer monitor.

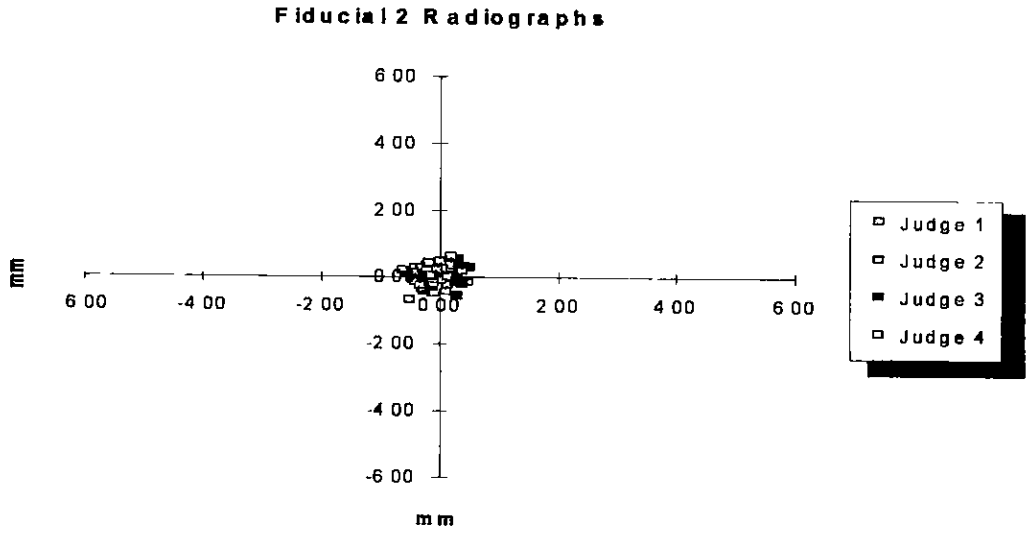


Figure 19 Error associated with the location of Fiducial 2 from all twenty five radiographs

Table 9.– Intra-Examiner Reliability Fiducial 2

Fiducial 2 x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.375	0.141	0.01	2.52	0.0188
2	0.566	0.320	0.03	-1.34	0.1933
3	0.359	0.129	0.01	-3.06	0.0055
4	0.363	0.132	0.02	-2.52	0.0187
1m	1.000	1.000	0.11	0.00	1.0000
4m	-1.000	1.000	0.16	2.07	177.7805
Fiducial 2 y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.131	0.017	0.00	1.28	0.2134
2	0.438	0.192	0.04	-0.32	0.7498
3	0.145	0.021	0.01	-2.22	0.0362
4	-0.174	0.030	-0.01	0.92	0.3661
1m	1.000	1.000	1.46	0.00	1.0000
4m	-0.570	0.325	0.29	2.07	1.2152

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

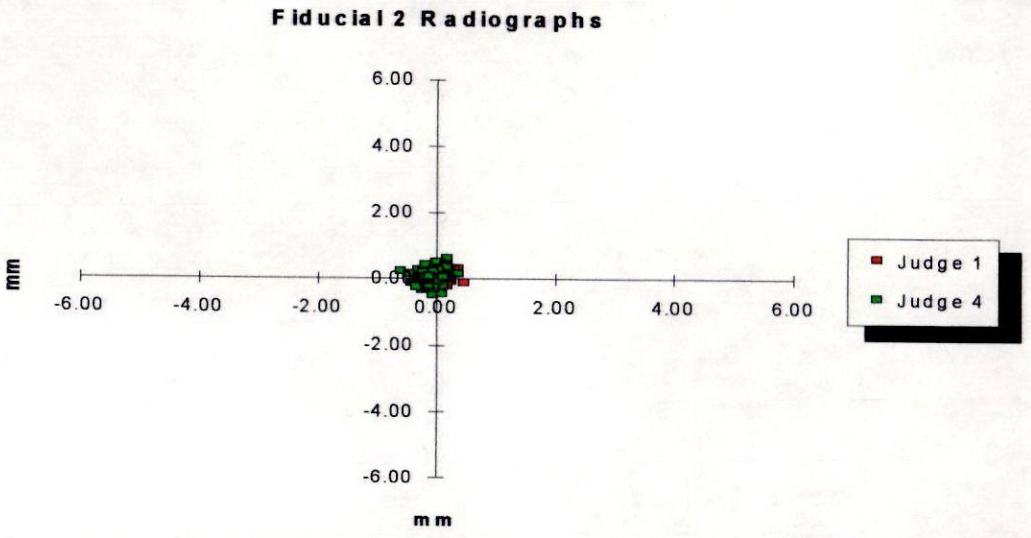


Figure 20. Error associated with the location of Fiducial 2 for Judge 1 and 4 from the radiographs.

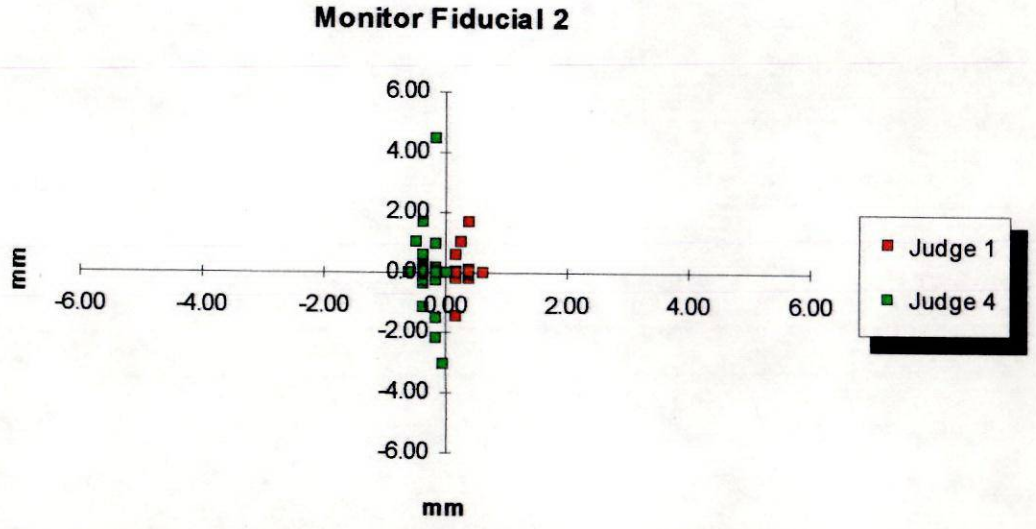


Figure 21. Error associated with the location of Fiducial 2 for Judge 1 and 4 from the computer monitor.

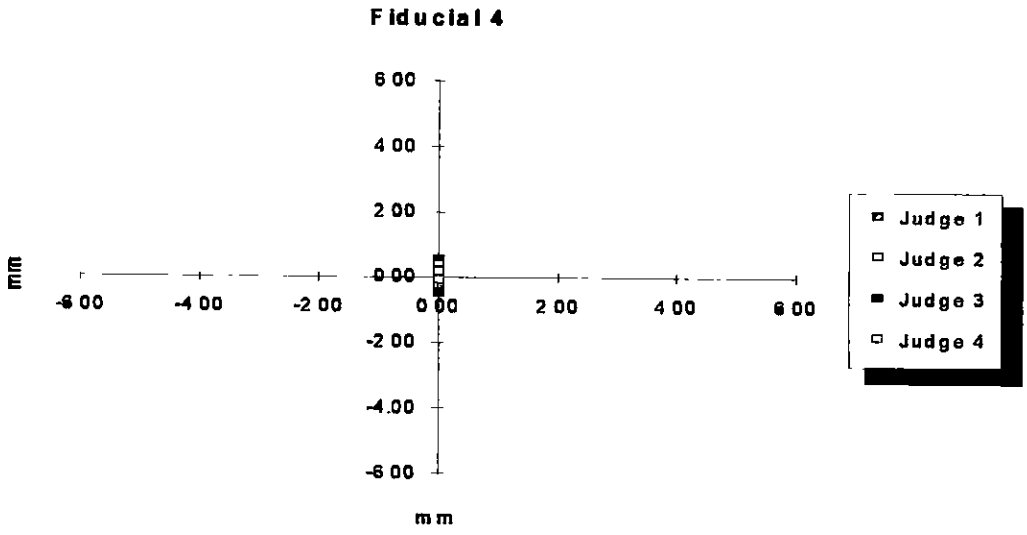


Figure 22. Error associated with the location of Fiducial 4 from all twenty five radiographs.

Table 10.-- Intra-Examiner Reliability Fiducial 4

Fiducial 4 x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1.00					
2.00					
3.00					
4.00					
1m	0.970	0.940	1.15	1.00	0.3293
4m	0.414	0.171	0.20	-0.96	0.3587
Fiducial 4 y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1.00	0.251	0.063	0.00	0.94	0.3570
2.00	0.285	0.081	0.01	-0.62	0.5405
3.00	-0.145	0.021	0.00	-4.30	0.0003
4.00	0.026	0.001	0.00	-3.44	0.0021
1m					
4m	0.537	0.288	0.08	1.43	0.1794

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

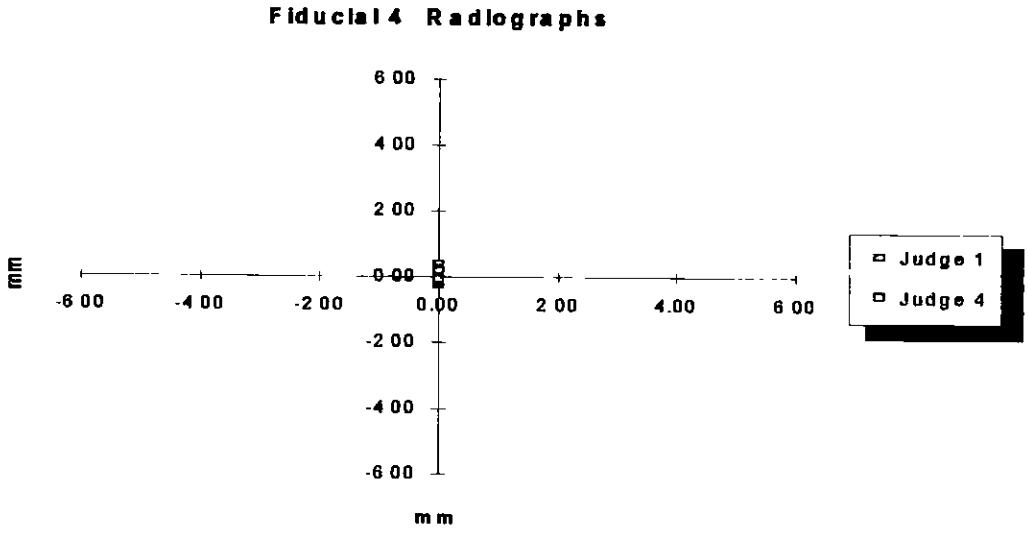


Figure 23. Error associated with the location of Fiducial 4 from Judge 1 and 4 from the radiographs.

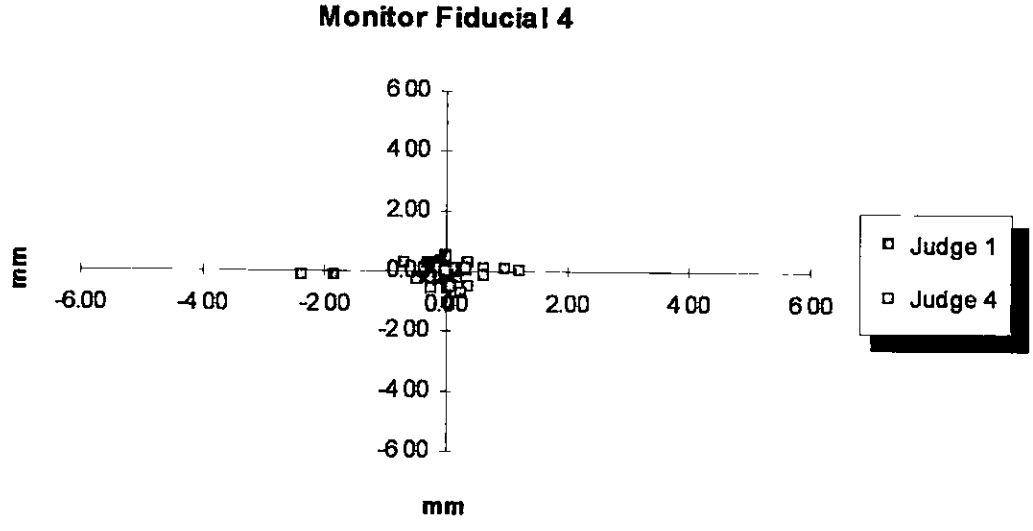


Figure 24. Error associated with the location of Fiducial 4 from Judge 1 and 4 from the computer monitor.

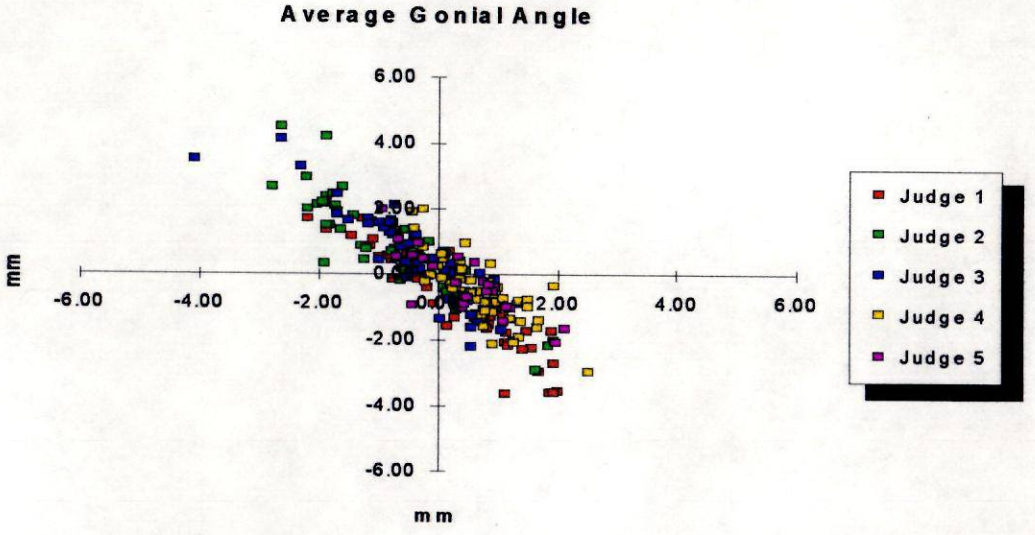


Figure 25. Error associated with the location of Gonial Angle from all twenty five radiographs.

Table 11.-- Intra-Examiner Reliability Gonial Angle

Gonial Angle x coordinate Alpha = 0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.995	0.991	67.98	0.95	0.3513
2	0.988	0.975	67.07	2.27	0.0328
3	0.984	0.969	67.71	2.15	0.0422
4	0.995	0.990	67.20	-0.57	0.5762
1m	0.995	0.989	72.32	0.37	0.7175
4m	0.994	0.989	68.74	3.92	0.0007
Gonial Angle y coordinate Alpha = 0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.979	0.959	61.22	-2.56	0.0172
2	0.979	0.959	61.22	-2.56	0.0172
3	0.967	0.936	65.41	-0.87	0.3955
4	0.987	0.974	63.45	1.74	0.0944
1m	0.990	0.981	69.22	0.23	0.8194
4m	0.990	0.980	67.30	-4.50	0.0002

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

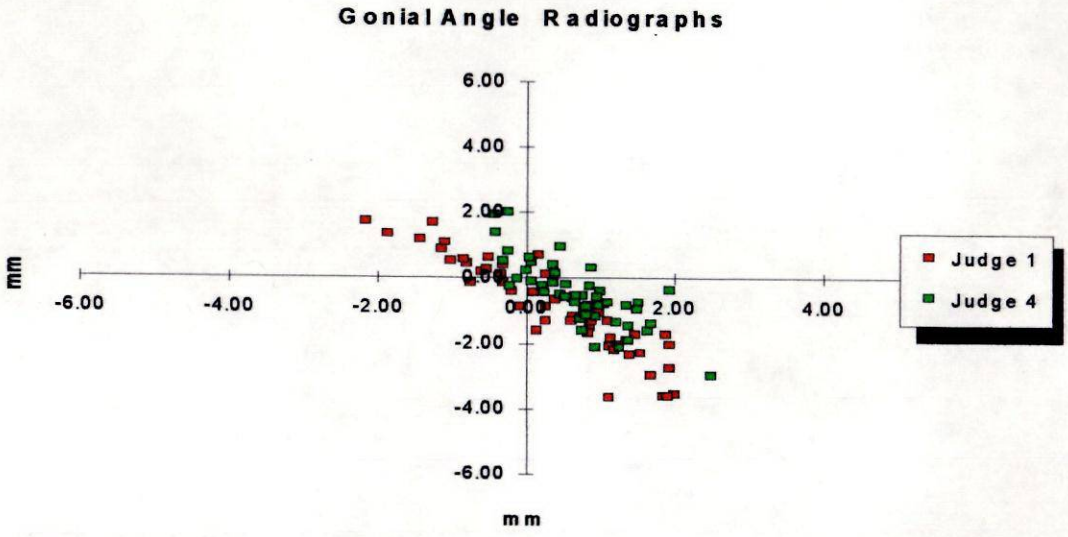


Figure 26. Error associated with the location of Gonial Angle for Judge 1 and 4 from the radiographs.

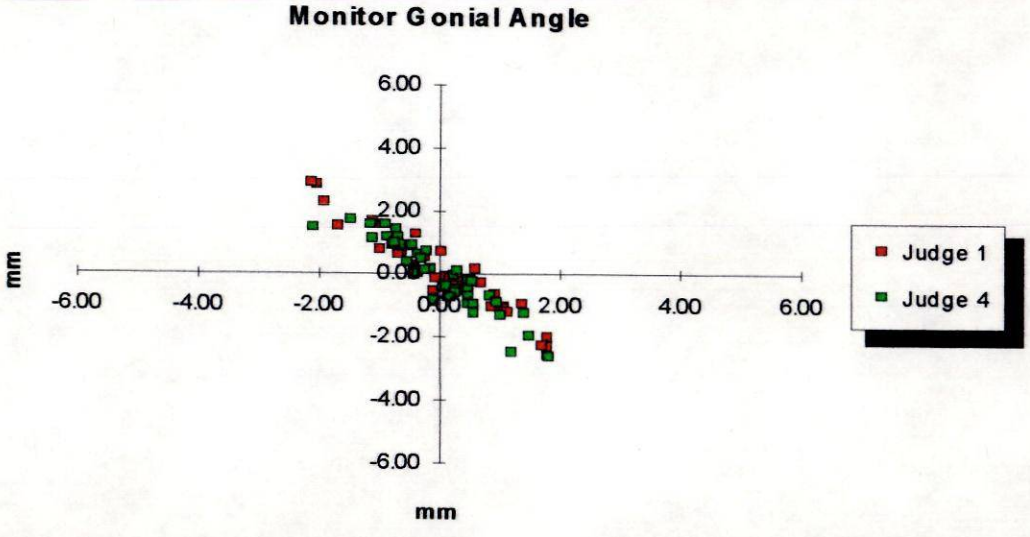


Figure 27. Error associated with the location of Gonial Angle for Judge 1 and 4 from the computer monitor.

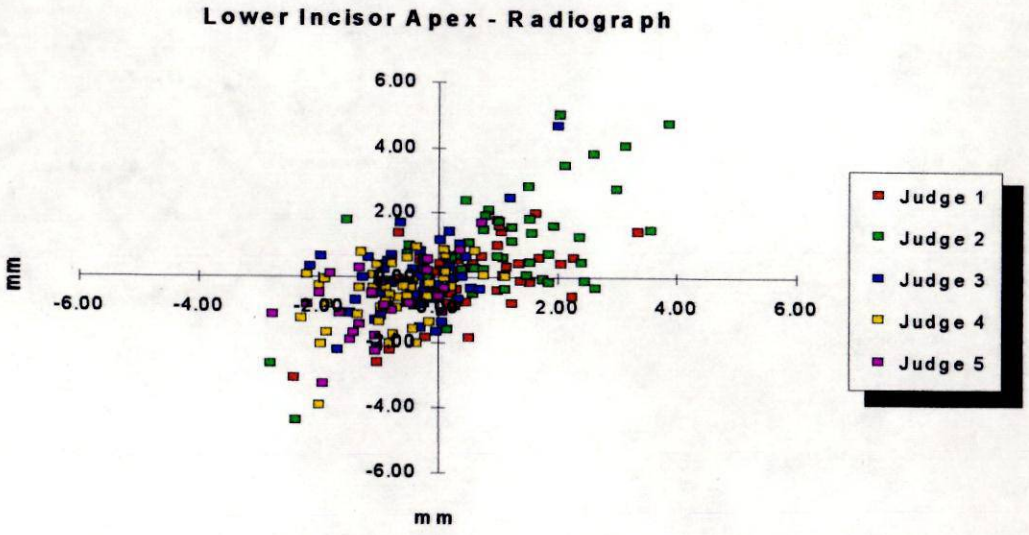


Figure 28. Error associated with the location of Lower Incisor Apex from all twenty five radiographs.

Table 12.-- Intra-Examiner Reliability Lower Incisor Apex

Lower Incisor Apex x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.954	0.910	7034.07	-0.98	0.3357
2	0.978	0.957	97.29	-1.78	0.0890
3	0.994	0.988	93.10	1.64	0.1140
4	0.995	0.990	91.08	0.69	0.4959
1m	0.965	0.931	91.33	1.28	0.2137
4m	0.984	0.969	95.12	-1.69	0.1060
Lower Incisor Apex y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.852	0.726	33.94	1.46	0.1573
2	0.951	0.904	70.03	-0.19	0.8477
3	0.988	0.976	74.50	3.70	0.0012
4	0.990	0.981	75.68	0.23	0.8226
1m	0.998	0.995	69.95	1.59	0.1259
4m	0.760	0.578	51.49	0.69	0.4991

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

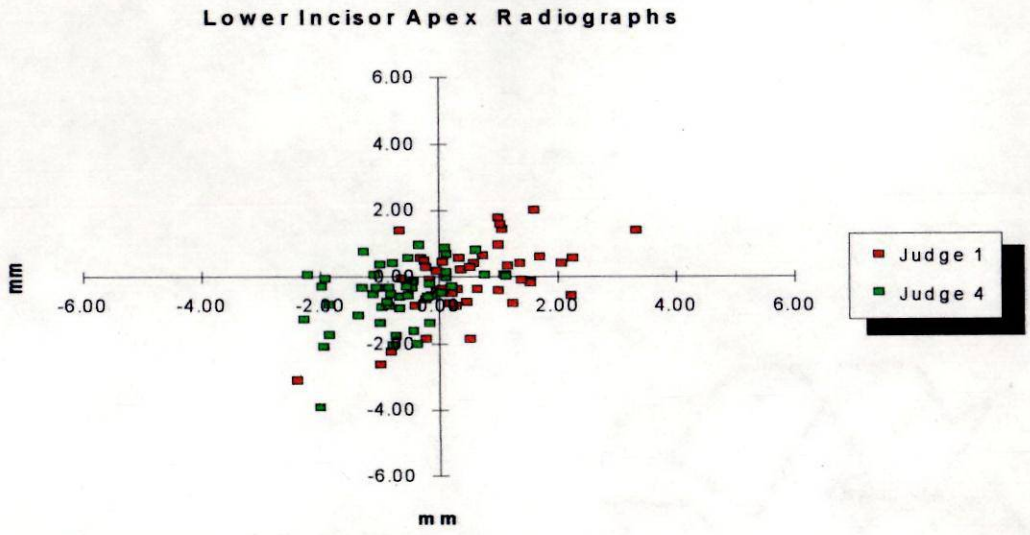


Figure 29. Error associated with the location of Lower Incisor Apex for Judge 1 and 4 from the radiographs.

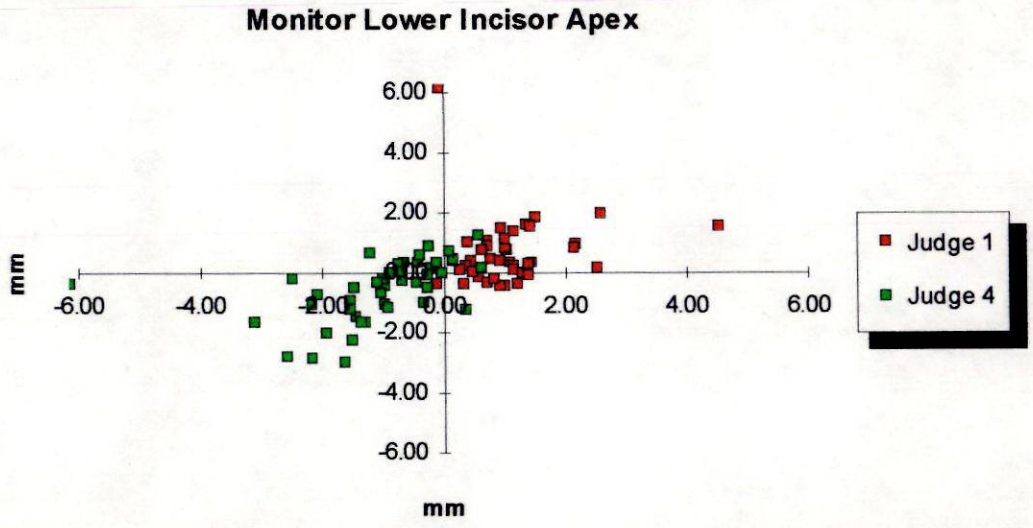


Figure 30. Error associated with the location of Lower Incisor Apex for Judge 1 and 4 from the computer monitor.

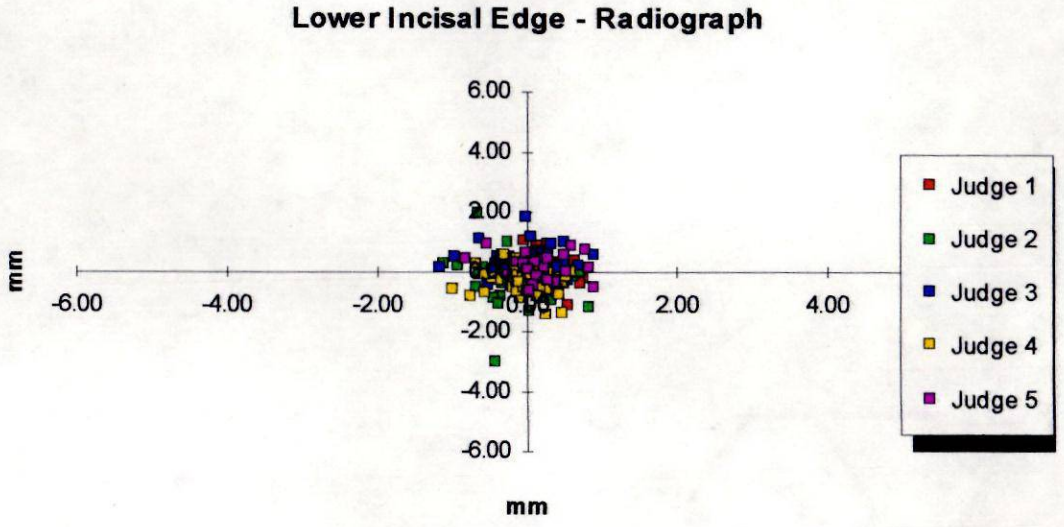


Figure 31. Error associated with the location of Lower Incisor Edge from all twenty five radiographs.

Table 13.-- Intra-Examiner Reliability Lower Incisor Edge

Lower Incisor Edge x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1.00	1.000	0.999	89.13	-1.25	0.2231
2.00	0.998	0.996	91.13	0.14	0.8910
3.00	0.999	0.997	92.33	0.20	0.8437
4.00	0.998	0.997	91.44	-0.39	0.6972
1m	1.000	0.999	96.29	0.53	0.6012
4m	0.999	0.998	95.71	0.81	0.4291
Lower Incisor Edge y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1.00	0.998	0.997	87.99	-0.52	0.6060
2.00	0.995	0.989	85.55	-0.48	0.6369
3.00	0.997	0.995	84.98	1.60	0.1228
4.00	0.998	0.997	87.42	-1.43	0.1668
1m	0.999	0.999	89.44	0.54	0.5948
4m	0.999	0.999	89.55	0.77	0.4515

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

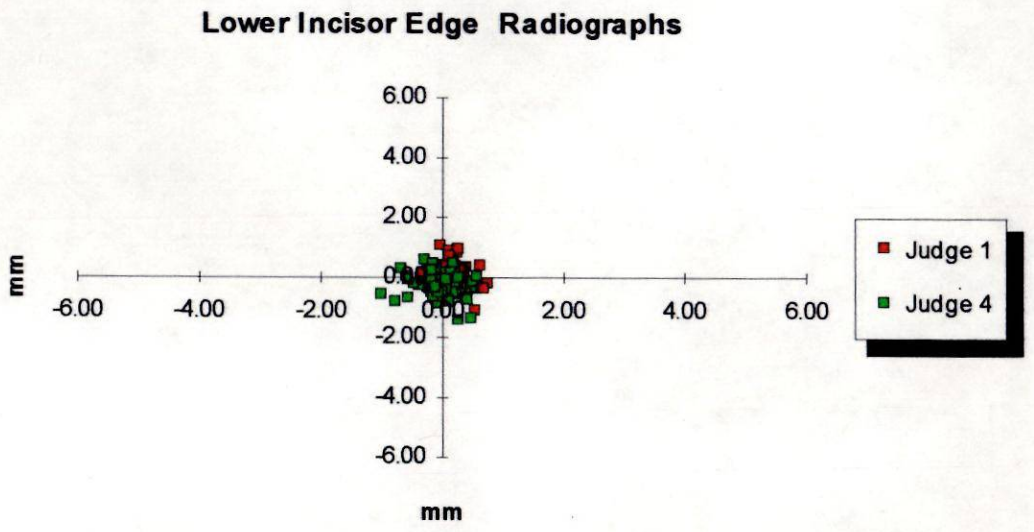


Figure 32. Error associated with the location of Lower Incisor Edge for Judge 1 and 4 from the radiographs.

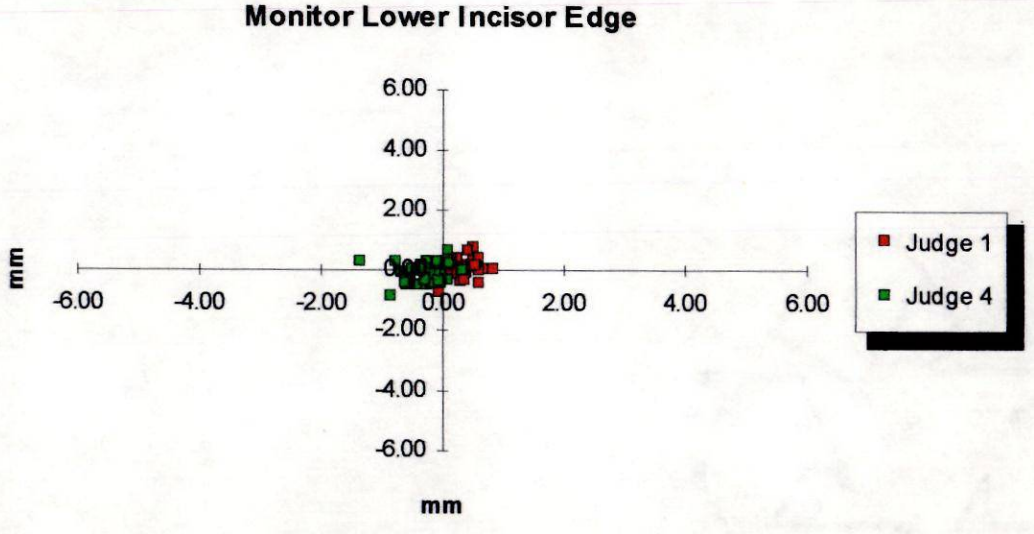


Figure 33. Error associated with the location of Lower Incisor Edge for Judge 1 and 4 from the computer monitor.

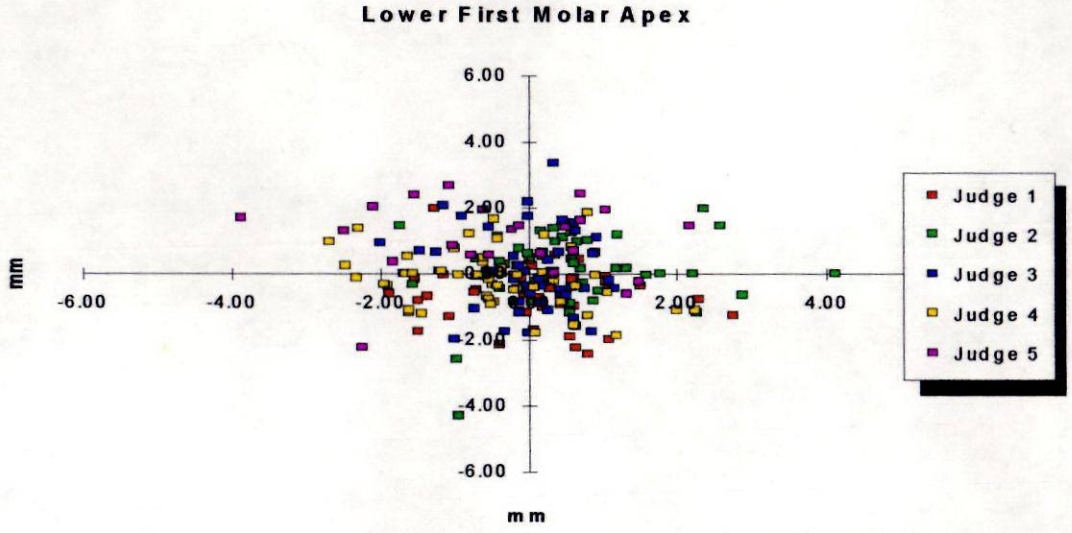


Figure 34. Error associated with the location of Lower First Molar Mesiobuccal Root Apex from all twenty five radiographs.

Table 14.-- Intra-Examiner Reliability Lower First Molar Mesiobuccal Root Apex

Lower First Molar MB Root Apex x coordinate Alpha=0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.943	0.888	1687.10	-0.95	0.3528
2	0.986	0.972	75.26	-2.49	0.0205
3	0.982	0.964	83.89	-1.22	0.2354
4	0.993	0.986	86.39	-2.91	0.0077
1m	0.998	0.997	699.17	0.76	0.4555
4m	0.997	0.994	85.95	3.51	0.00
Lower First Molar Mesiobuccal Root Apex y coordinate Alpha=0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.970	0.940	48.84	1.98	0.0592
2	0.975	0.950	63.20	0.63	0.5319
3	0.991	0.982	70.63	2.82	0.0100
4	0.993	0.986	65.02	2.89	0.0080
1m	0.992	0.983	73.49	-0.62	0.5393
4m	0.977	0.955	67.27	0.80	0.43

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

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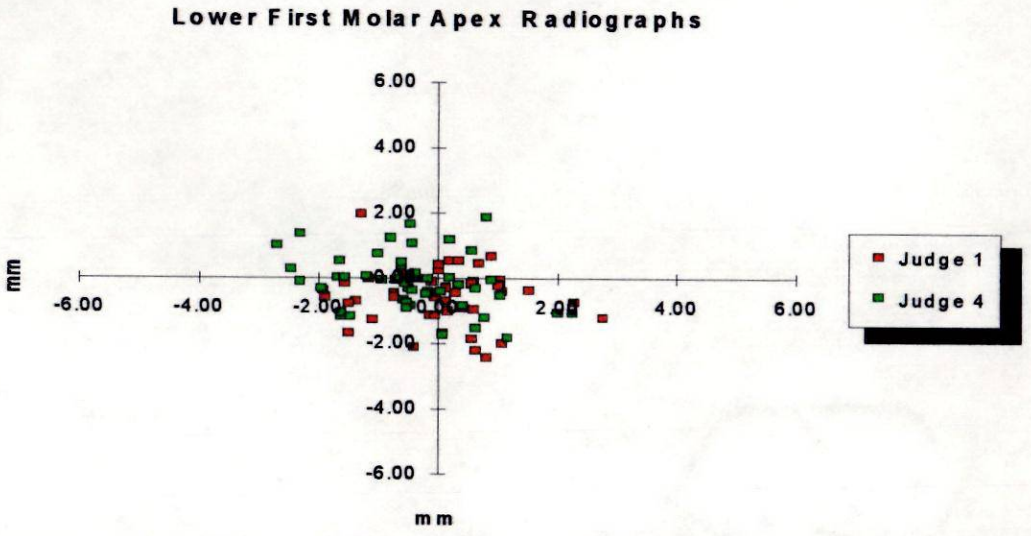


Figure 35. Error associated with the location of Lower First Molar Mesiobuccal Root Apex for Judge 1 and 4 from the radiographs.

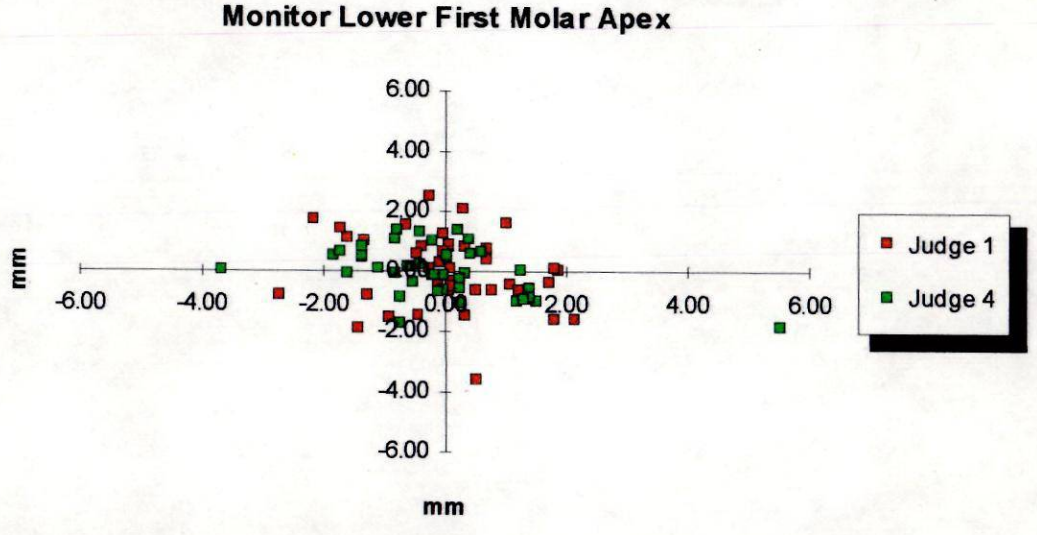


Figure 36. Error associated with the location of Lower First Molar Mesiobuccal Root Apex for Judge 1 and 4 from the computer monitor.

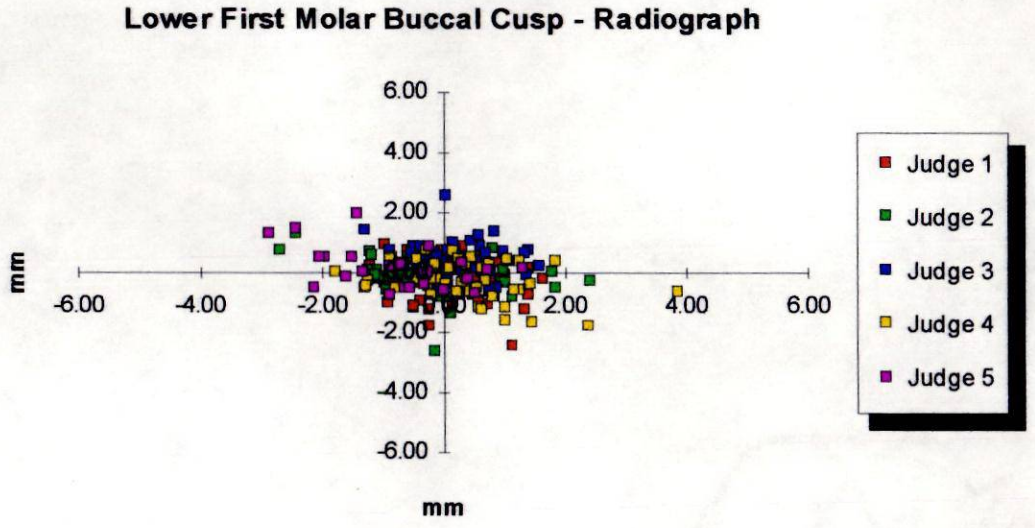


Figure 37. Error associated with the location of Lower First Molar Mesiobuccal Cusp from all twenty five radiographs.

Table 15.-- Intra-Examiner Reliability Lower First Molar Mesiobuccal Cusp

Lower First Molar Mesiobuccal Cusp x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.997	0.993	83.08	-0.14	0.8921
2	0.991	0.982	85.20	-0.62	0.5439
3	0.998	0.995	84.80	-0.03	0.9741
4	0.994	0.988	85.13	-1.59	0.1254
1m	0.999	0.999	88.43	1.06	0.3017
4m	0.996	0.992	81.73	-0.53	0.6003
Lower First Molar Mesiobuccal Cusp y coordinate Alpha =0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.996	0.992	78.08	-0.07	0.9476
2	0.995	0.991	76.55	0.55	0.5859
3	0.997	0.993	77.85	-0.33	0.7466
4	0.995	0.991	76.77	2.95	0.0070
1m	0.999	0.999	75.61	-0.72	0.4820
4m	0.463	0.214	67.03	-0.98	0.3383

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

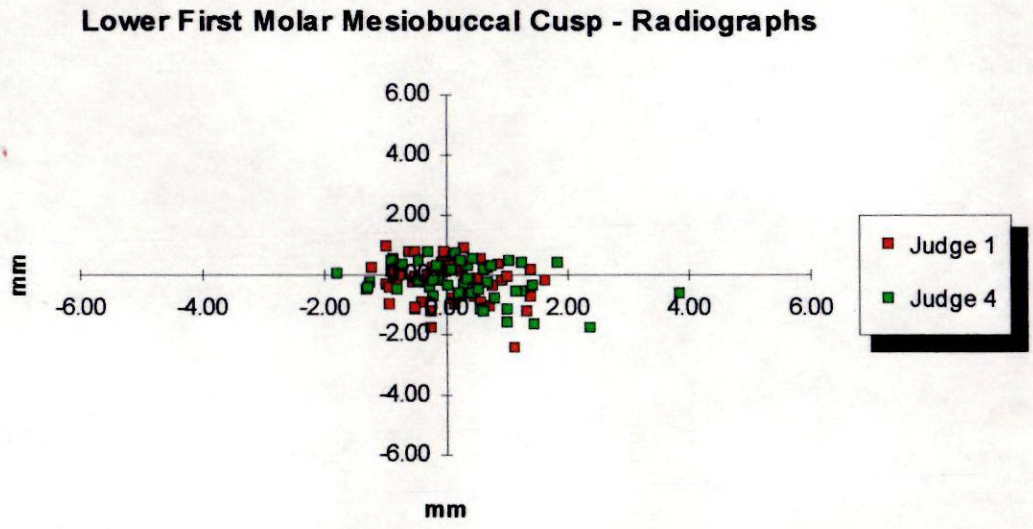


Figure 38. Error associated with the location of Lower First Molar Mesiobuccal Cusp for Judge 1 and 4 from the radiographs.

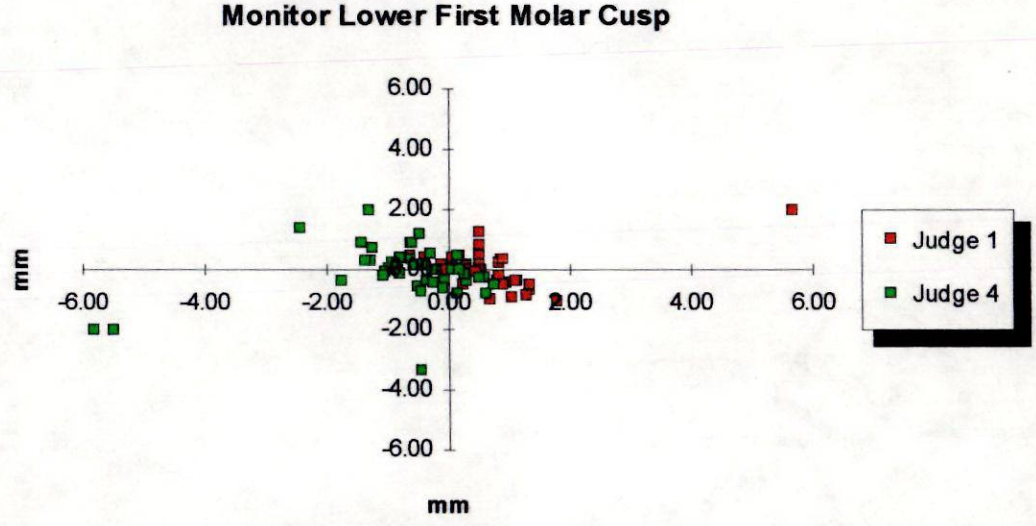


Figure 39. Error associated with the location of Lower First Molar Mesiobuccal Cusp for Judge 1 and 4 from the computer monitor.

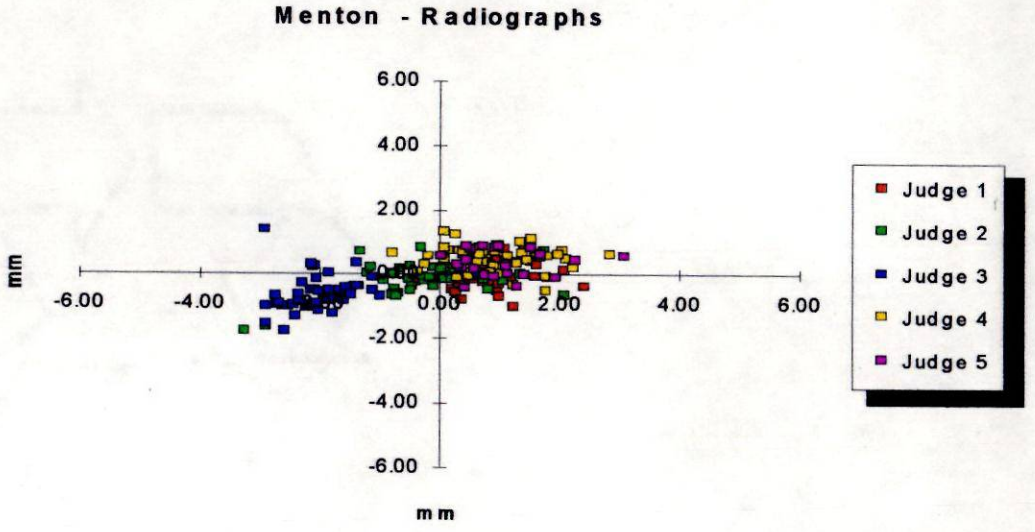


Figure 40. Error associated with the location of Menton from all twenty five radiographs.

Table 16. -- Intra-Examiner Reliability Menton

Menton x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.999	0.997	88.65	-1.85	0.0777
2	0.987	0.975	95.27	-2.59	0.0163
3	0.995	0.991	97.83	-0.57	0.5712
4	0.994	0.988	90.08	-0.96	0.3477
1m	0.999	0.999	100.18	-0.62	0.5441
4m	0.999	0.997	973.69	-1.53	0.1396
Menton y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.997	0.995	38.72	1.04	0.3094
2	0.990	0.980	39.21	-0.11	0.9170
3	0.994	0.988	37.97	1.71	0.1018
4	0.998	0.997	40.23	1.74	0.0957
1m	0.947	0.897	48.02	1.02	0.3208
4m	0.983	0.966	39.09	1.35	0.1917

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

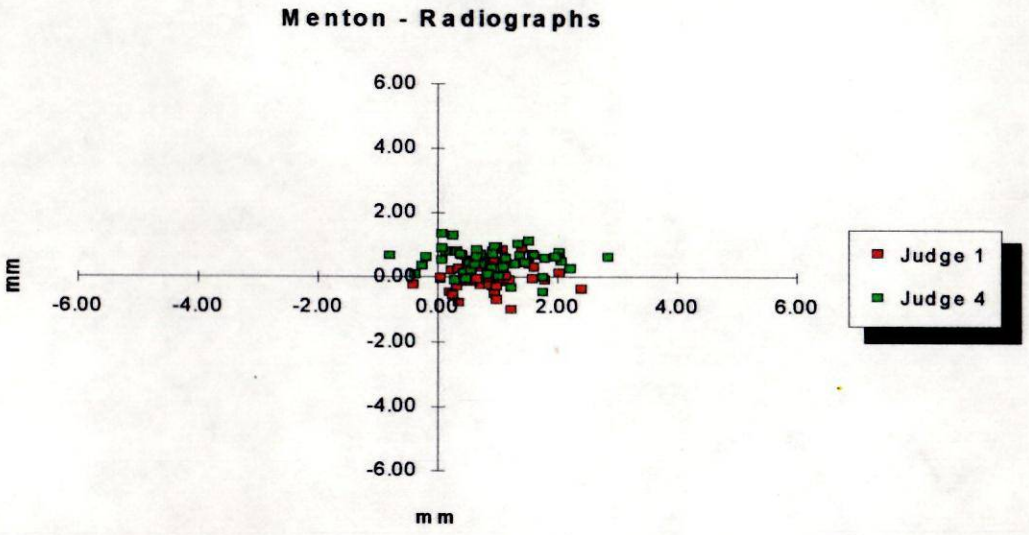


Figure 41. Error associated with the location of Menton for Judge 1 and 4 from the radiographs.

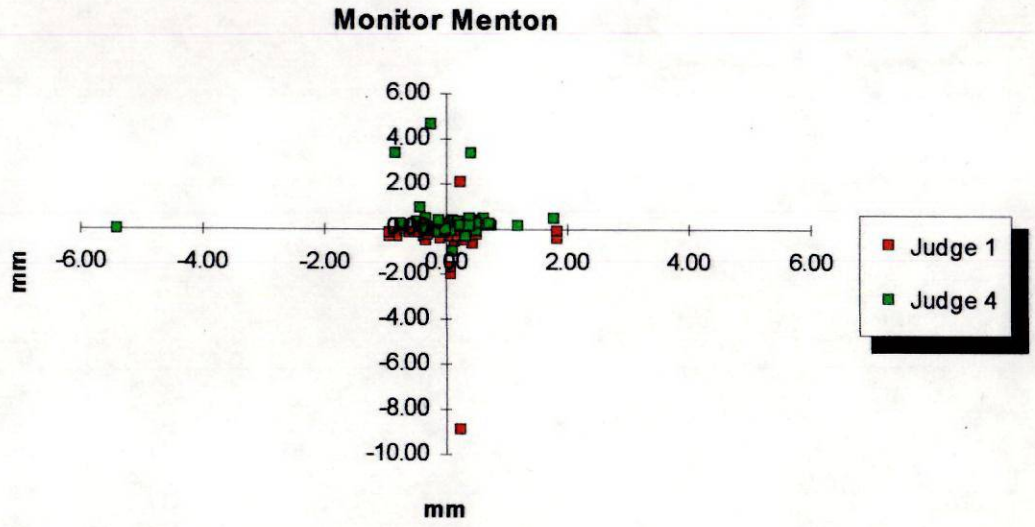


Figure 42. Error associated with the location of Menton for Judge 1 and 4 from the computer monitor.

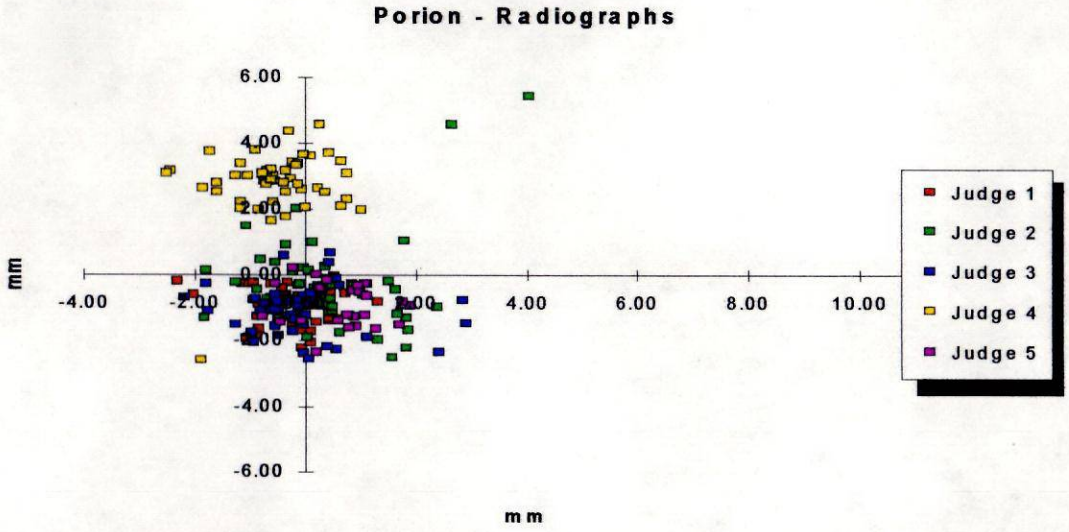


Figure 43. Error associated with the location of Mechanical Porion from all twenty five radiographs.

Table 17.-- Intra-Examiner Reliability Mechanical Porion

Porion x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.998	0.995	64.75	-0.59	0.5578
2	0.993	0.986	64.47	-7.44	0.0000
3	0.980	0.961	63.23	-1.98	0.0597
4	0.994	0.989	68.97	-1.07	0.2937
1m					
4m					
Porion y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.999	0.997	140.19	0.49	0.6267
2	0.981	0.963	135.08	-0.21	0.8373
3	0.997	0.994	145.20	2.16	0.0413
4	0.992	0.984	134.82	-1.04	0.3101
1m					
4m					

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

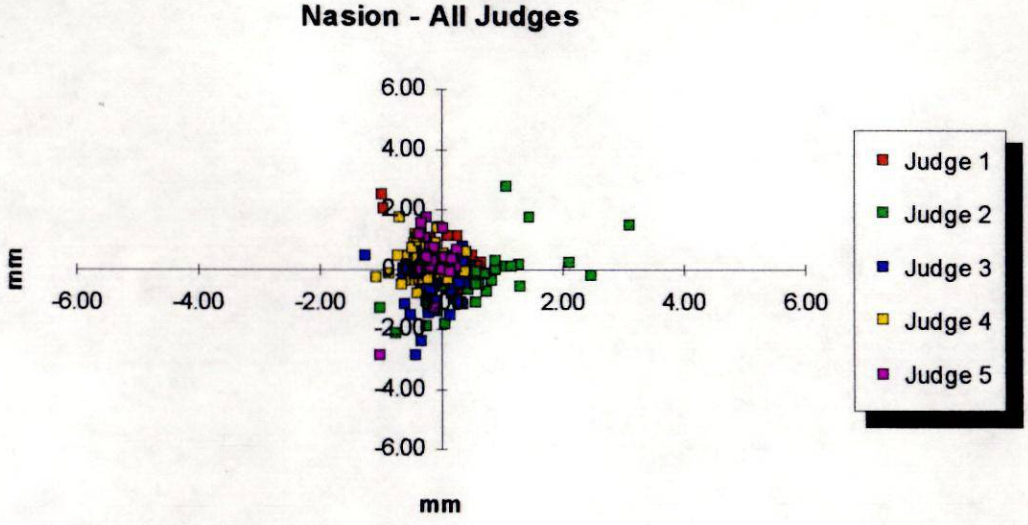


Figure 44. Error associated with the location of Nasion from all twenty five radiographs.

Table 18.-- Intra-Examiner Reliability Nasion

Nasion x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	1.000	0.999	69.2641	0.2378	0.8140
2	0.989	0.978	67.0842	0.0970	0.9235
3	0.998	0.996	68.5025	0.7367	0.4687
4	0.999	0.998	68.9375	-0.8483	0.4047
1m	1.000	0.999	78.2778	-1.2127	0.2381
4m	0.999	0.998	72.5604	2.2681	0.0335
Nasion y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.999	0.998	157.6473	0.5342	0.5981
2	0.994	0.989	145.7196	-1.3197	0.1994
3	0.999	0.997	152.7494	0.0624	0.9508
4	0.999	0.997	158.4681	-2.5511	0.0175
1m	0.998	0.997	142.6772	-1.6199	0.1195
4m	0.999	0.997	146.8220	0.5952	0.5578

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

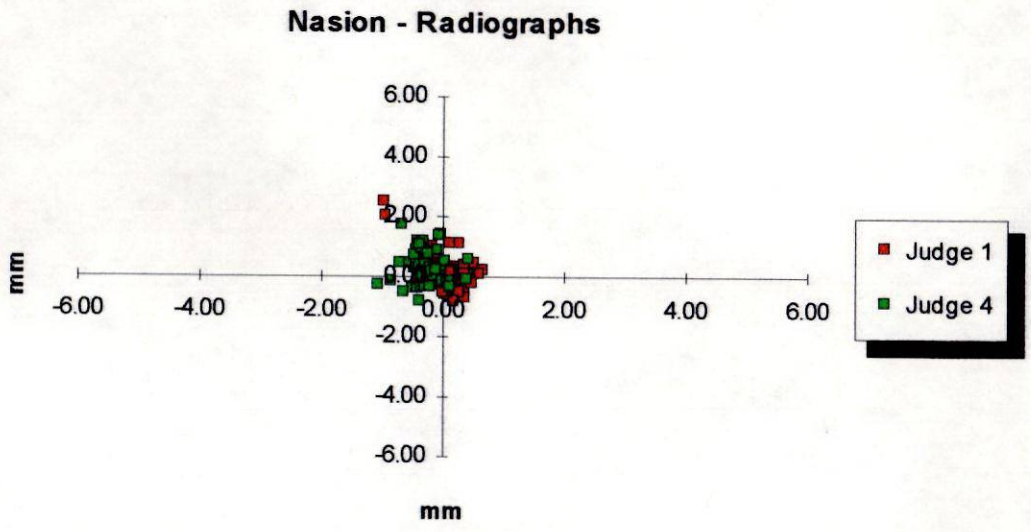


Figure 45. Error associated with the location of Nasion for Judge 1 and 4 from the radiographs.

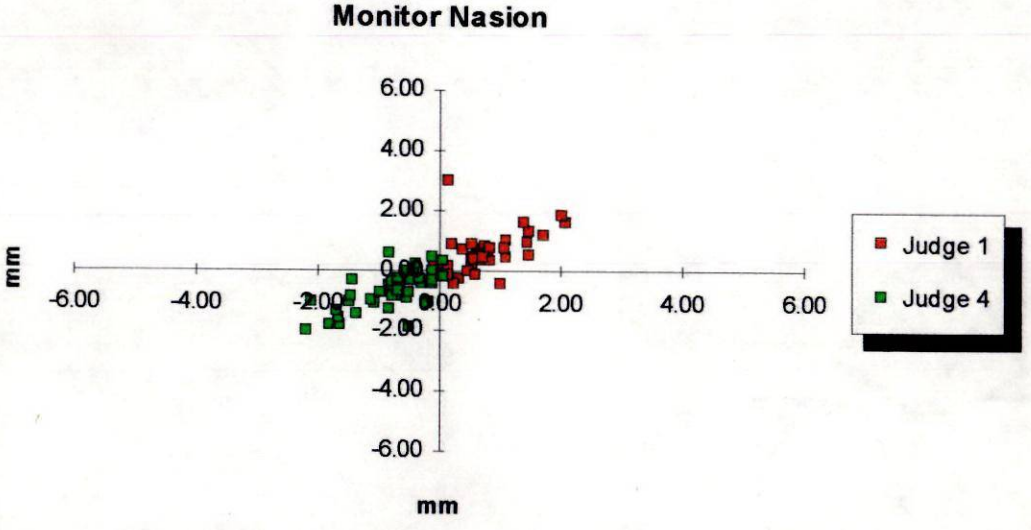


Figure 46. Error associated with the location of Nasion for Judge 1 and 4 from the computer monitor.

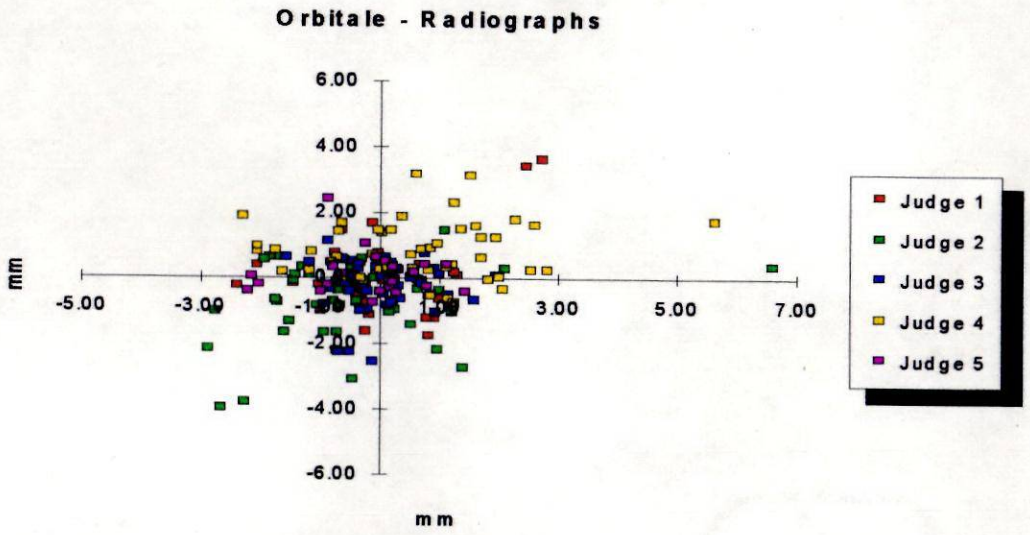


Figure 47. Error associated with the location of Orbitale from all twenty five radiographs.

Table 19.-- Intra-Examiner Reliability Orbitale

Orbitale x coordinate Alpha = 0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.997	0.994	62.57	0.83	0.4170
2	0.977	0.954	60.38	-1.65	0.1118
3	0.994	0.988	60.64	-0.49	0.6258
4	0.962	0.926	65.16	0.31	0.7557
1m	0.998	0.996	77.42	-1.04	0.3093
4m	0.984	0.969	65.11	-4.33	0.0003
Orbitale y coordinate Alpha = 0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.998	0.996	116.57	0.15	0.8812
2	0.994	0.988	115.92	-0.70	0.4892
3	0.996	0.992	123.56	0.57	0.5711
4	0.997	0.994	126.69	-3.07	0.0053
1m	0.999	0.997	122.46	0.55	0.5884
4m	0.996	0.992	126.35	0.00	0.9986

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

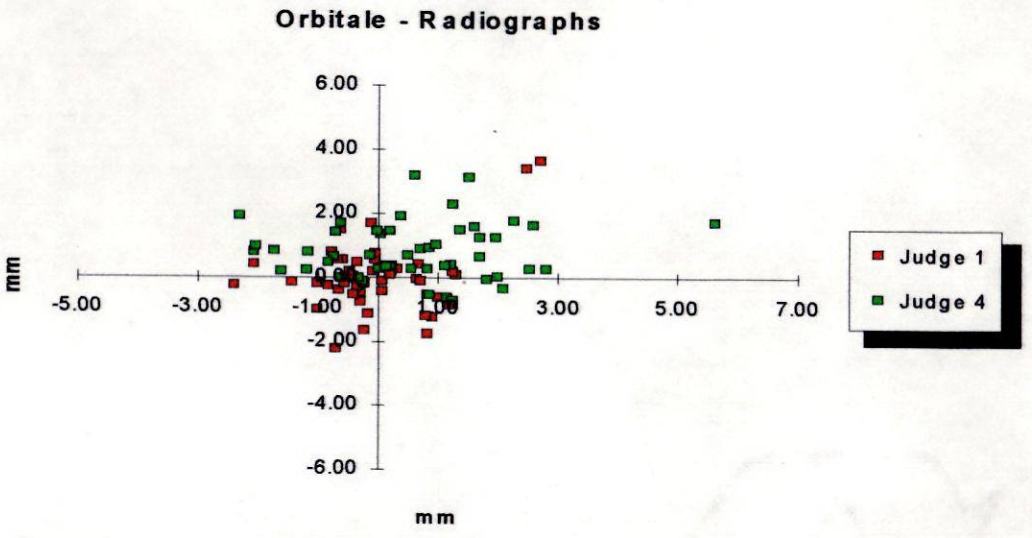


Figure 48. Error associated with the location of Orbitale for Judge 1 and 4 from the radiographs.

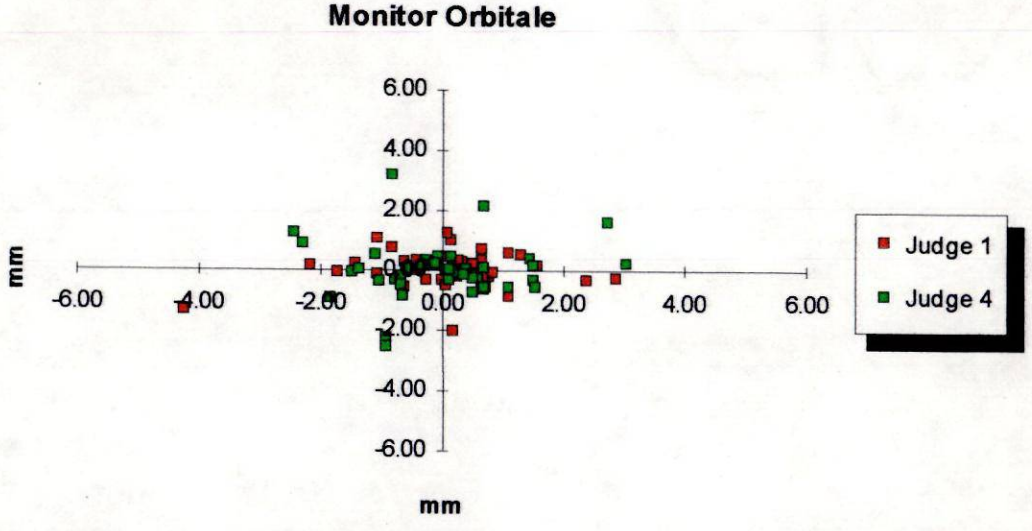


Figure 49. Error associated with the location of Orbitale for Judge 1 and 4 from the computer monitor.

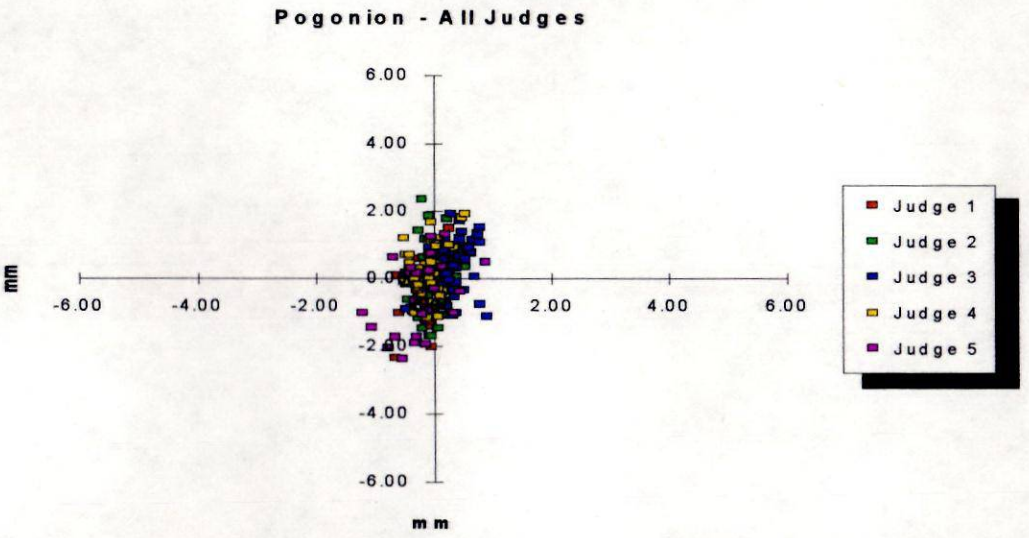


Figure 50. Error associated with the location of Pogonion from all twenty five radiographs.

Table 20.-- Intra-Examiner Reliability Pogonion

Pogonion x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	1.000	0.999	94.51	-2.32	0.0292
2	0.999	0.999	98.16	2.19	0.0389
3	0.999	0.999	100.03	1.80	0.0856
4	0.999	0.999	93.88	1.07	0.2955
1m	1.000	0.999	107.73	0.09	0.9301
4m	0.999	0.999	104.46	-2.07	0.0509
Pogonion y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.992	0.985	54.07	-0.74	0.4678
2	0.973	0.947	41.82	1.12	0.2738
3	0.993	0.985	41.93	0.44	0.6634
4	0.993	0.986	62.20	0.20	0.8447
1m	0.840	0.706	30.70	1.01	0.3241
4m	0.912	0.831	53.49	1.14	0.2652

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

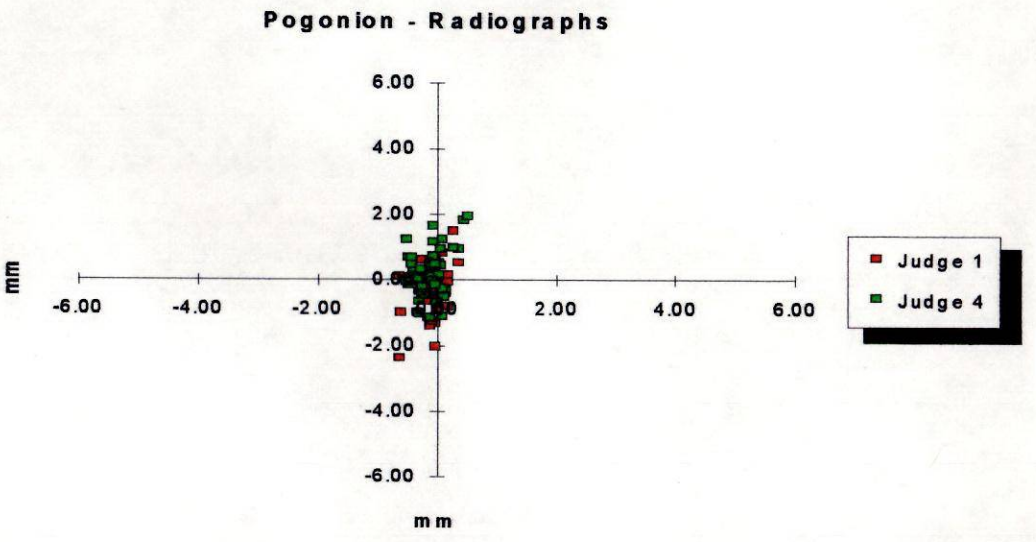


Figure 51. Error associated with the location of Pogonion for Judge 1 and 4 from the radiographs.

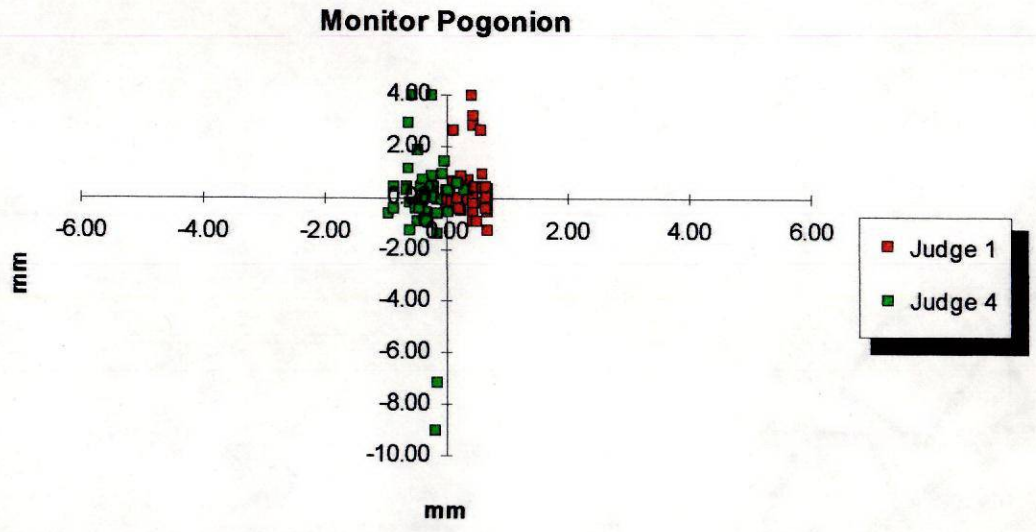


Figure 52. Error associated with the location of Pogonion for Judge 1 and 4 from the computer monitor.

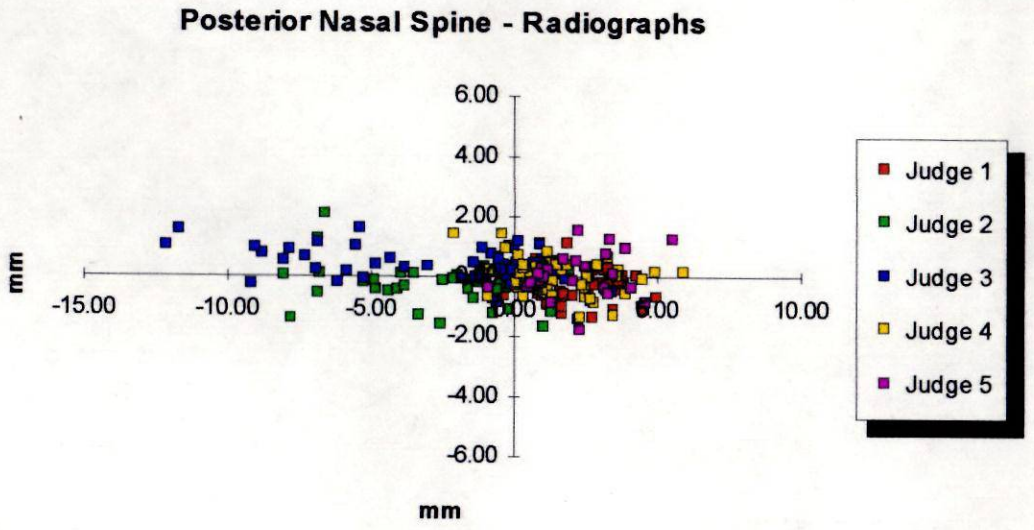


Figure 53. Error associated with the location of Posterior Nasal Spine from all twenty five radiographs.

Table 21.-- Intra-Examiner Reliability Posterior Nasal Spine

PNS x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.995	0.989	68.02	2.09	0.0477
2	0.928	0.862	79.10	1.40	0.1740
3	0.857	0.735	66.93	4.94	0.0001
4	0.985	0.971	66.69	1.06	0.2990
1m	0.997	0.993	70.02	-0.90	0.3763
4m	0.993	0.986	69.47	0.72	0.4806
PNS y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.998	0.995	102.42	-1.09	0.2869
2	0.997	0.995	101.90	3.63	0.0013
3	0.998	0.996	105.93	-1.59	0.1260
4	0.998	0.997	98.62	1.32	0.1989
1m	0.997	0.993	93.60	-0.14	0.8867
4m	0.997	0.994	99.15	-0.85	0.4046

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

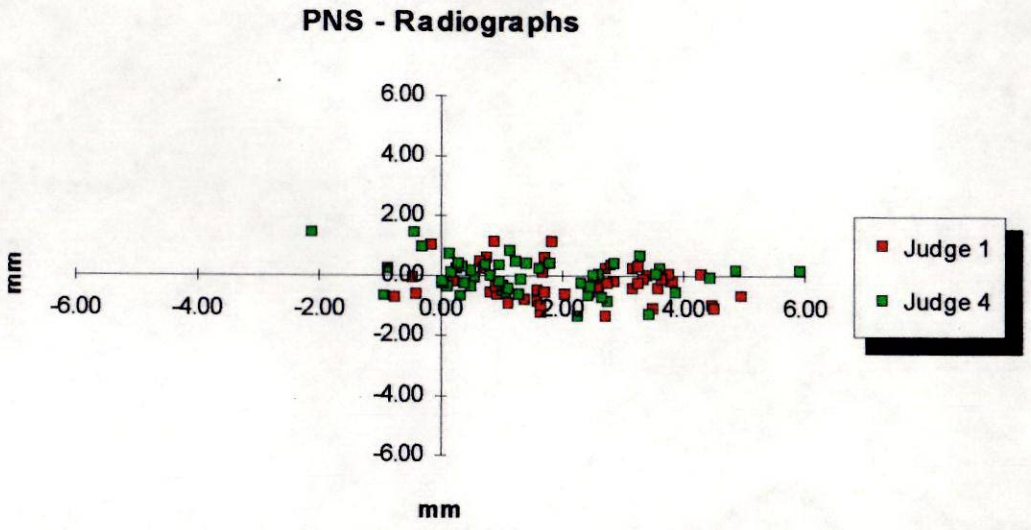


Figure 54. Error associated with the location of Posterior Nasal Spine for Judge 1 and 4 from the radiographs.

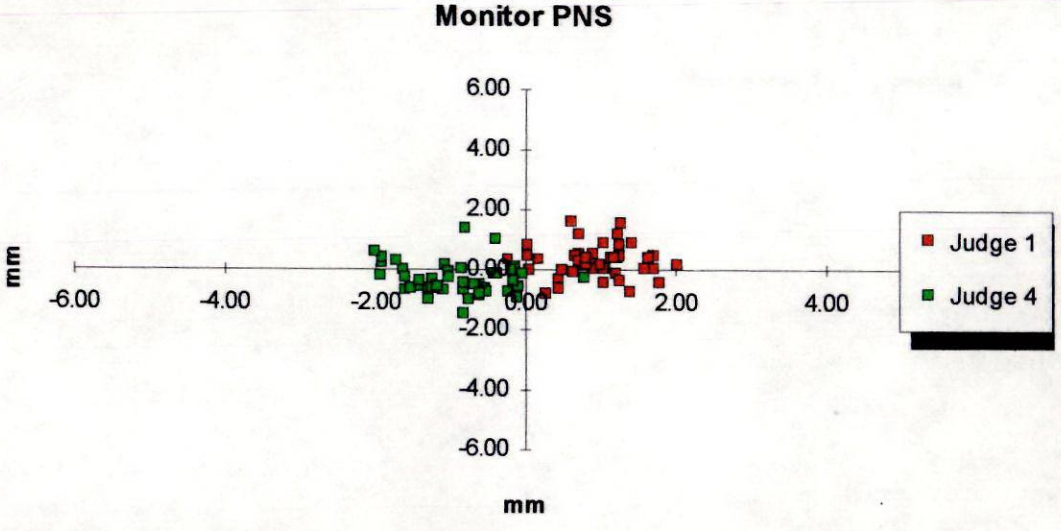


Figure 55. Error associated with the location of Posterior Nasal Spine Judge 1 and 4 from the computer monitor.

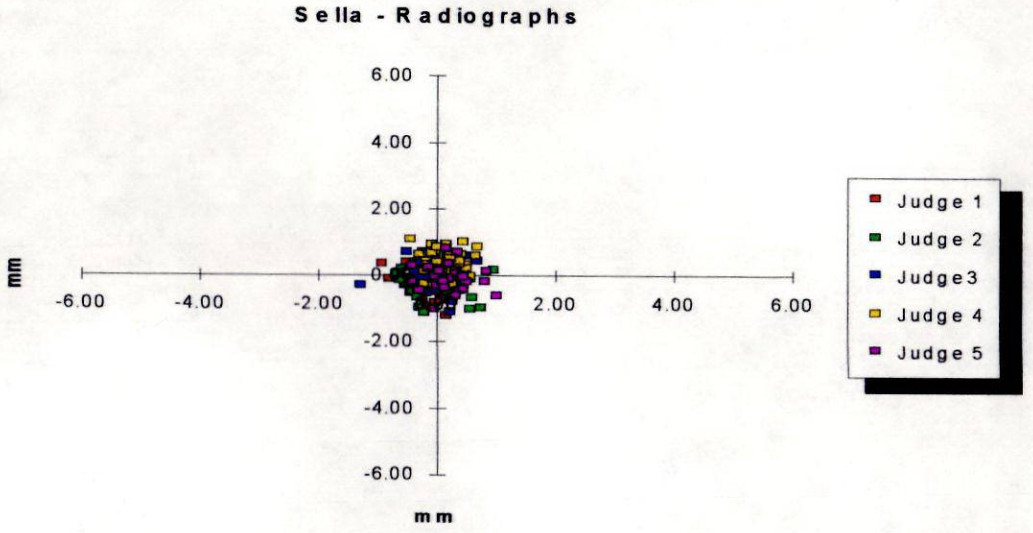


Figure 56. Error associated with the location of Sella from all twenty five radiographs.

Table 22.-- Intra-Examiner Reliability Sella

Sella x coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.999	0.998	59.75	1.05	0.3032
2	0.998	0.997	62.08	-3.94	0.0006
3	0.997	0.995	57.2346	0.62	0.5417
4	0.999	0.998	60.76	0.05	0.9574
1m	1.000	0.999	51.97	1.43	0.1724
4m	1.000	1.000	1363.60	1.52	0.1436
Sella y coordinate Alpha = 0.05					
Judge	**r**	**r^2**	**Pooled Variance**	**t**	**P(T<=t) two-tail**
1	0.999	0.998	170.16	-0.33	0.7422
2	0.999	0.999	167.65	-0.75	0.4620
3	0.999	0.998	174.62	-0.80	0.4341
4	1.000	0.999	168.31	-4.28	0.0003
1m	1.000	1.000	204.17	0.47	0.6432
4m	1.000	1.000	1840.99	2.29	0.0316

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

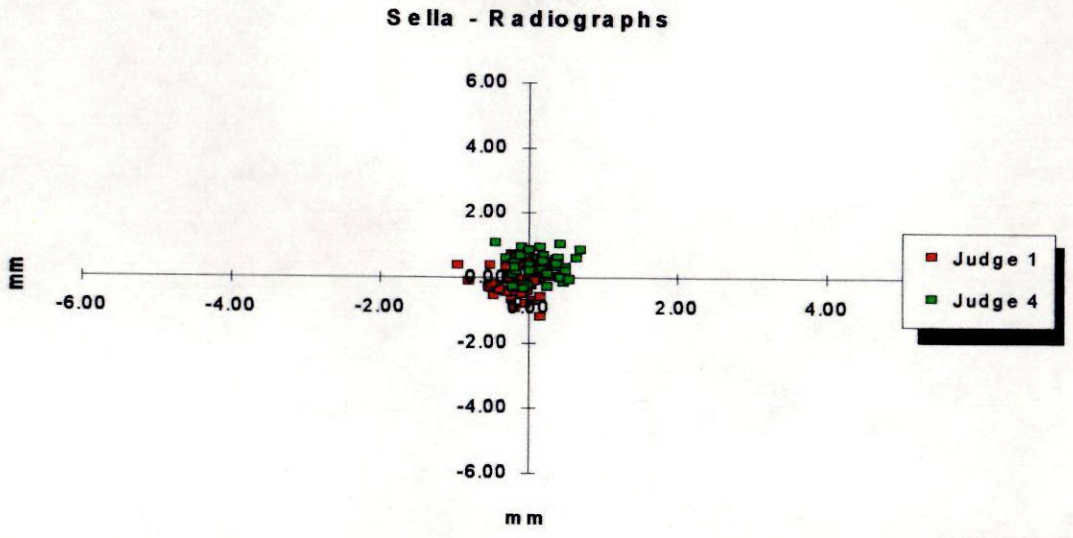


Figure 57. Error associated with the location of Sella for Judge 1 and 4 from the radiographs.

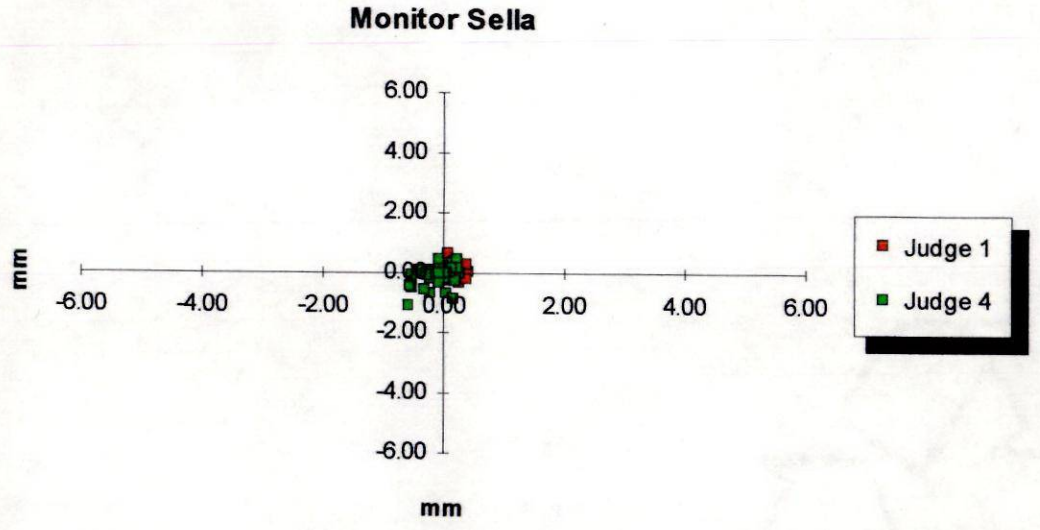


Figure 58. Error associated with the location of Sella for Judge 1 and 4 from the computer monitor.

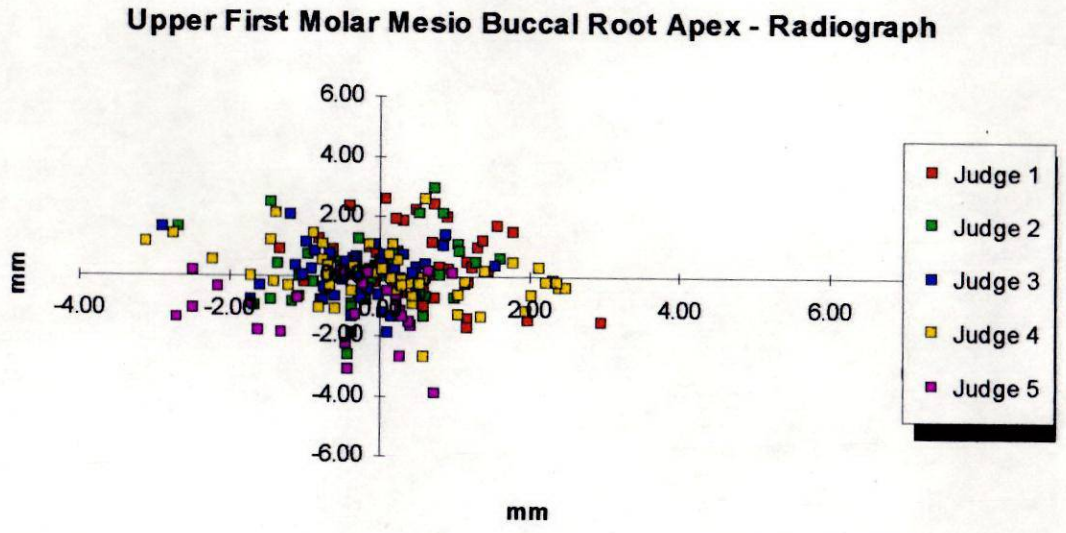


Figure 59. Error associated with the location of Upper First Molar Mesiobuccal Root Apex from all twenty five radiographs.

Table 23.-- Intra-Examiner Reliability Upper First Molar Mesiobuccal Root Apex

Upper First Molar Mesiobuccal Root Apex x coordinate Alpha =0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	1.000	0.999	738.71	1.82	0.0812
2	0.976	0.953	79.93	-2.41	0.0242
3	0.990	0.981	75.71	-0.77	0.4483
4	0.983	0.966	81.48	-2.18	0.0393
1m	1.000	0.999	785.45	-0.73	0.4706
4m	0.996	0.993	73.00	-1.16	0.2575
Upper First Molar Mesiobuccal Root Apex y coordinate Alpha =0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.996	0.993	178.75	0.59	0.5614
2	0.981	0.961	78.24	-0.41	0.6840
3	0.993	0.987	79.03	0.03	0.9782
4	0.992	0.983	74.93	3.16	0.0042
1m	0.997	0.995	183.67	-0.48	0.6335
4m	0.272	0.074	25.97	-1.12	0.2729

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

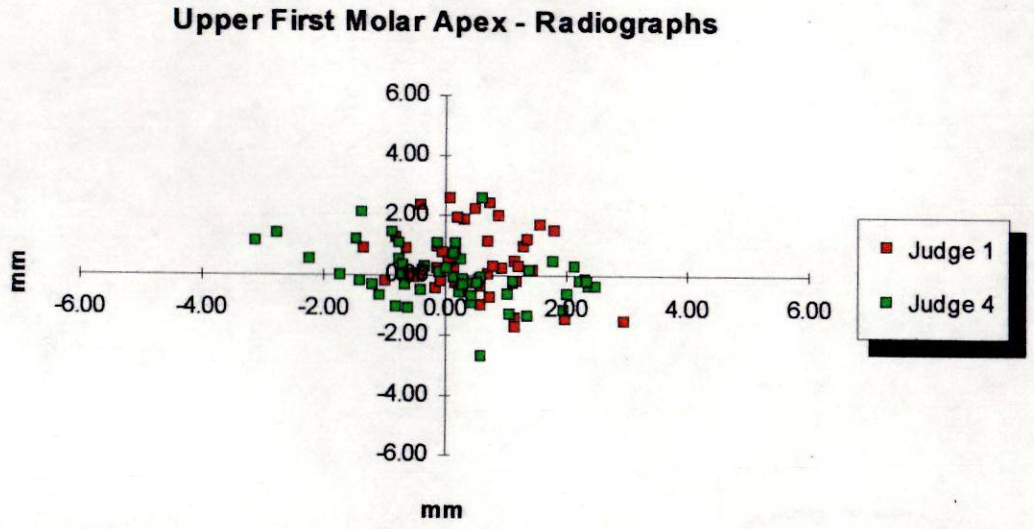


Figure 60. Error associated with the location of Upper First Molar Mesiobuccal Root Apex for Judge 1 and 4 from the radiographs.

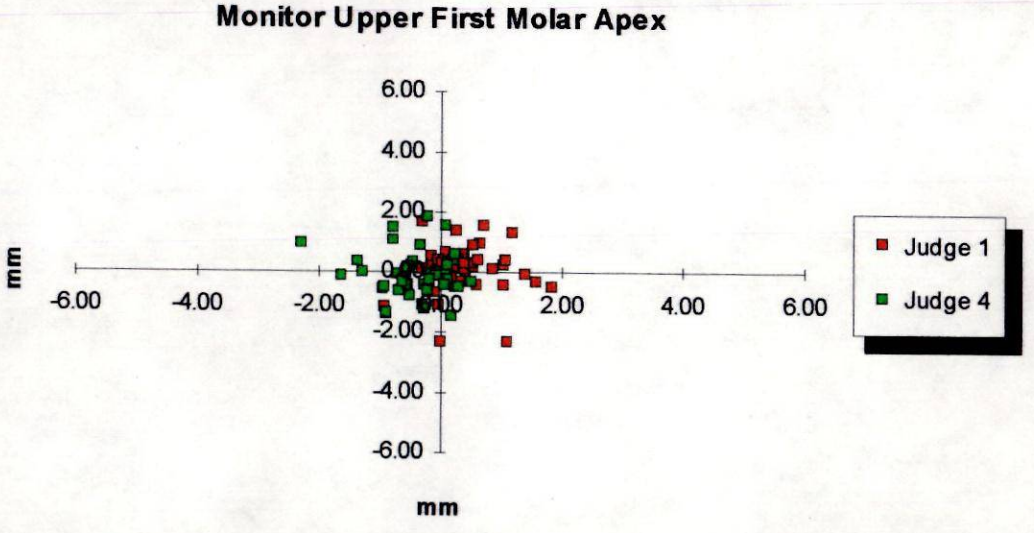


Figure 61. Error associated with the location of Upper First Molar Mesiobuccal Root Apex for Judge 1 and 4 from the computer monitor.

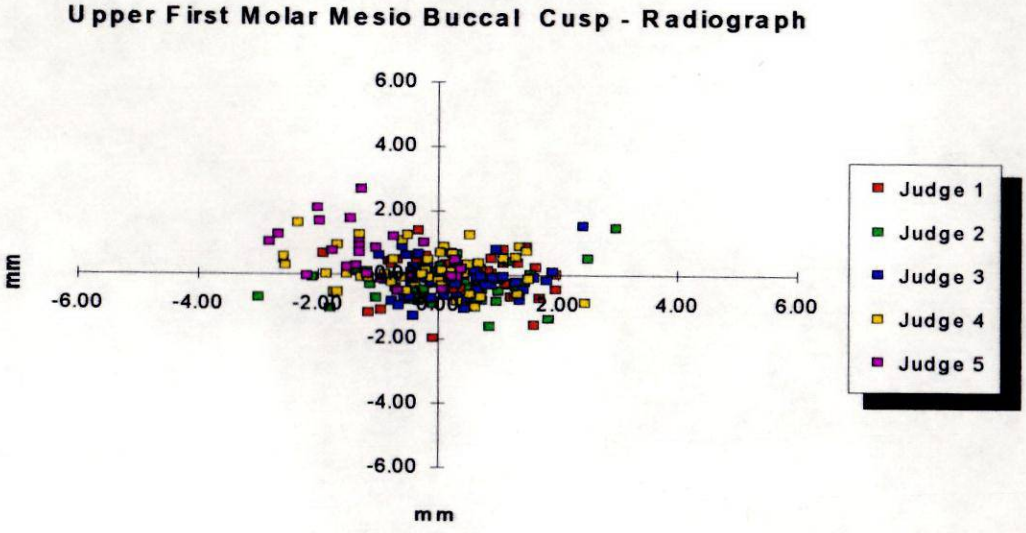


Figure 62. Error associated with the location of Upper First Molar Mesiobuccal Cusp from all twenty five radiographs.

Table 24.-- Intra-Examiner Reliability Upper First Molar Mesiobuccal Cusp

Upper First Molar Mesiobuccal Cusp x coordinate Alpha=0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.997	0.994	77.17	-0.05	0.9643
2	0.985	0.971	77.87	-1.14	0.2666
3	0.994	0.988	78.87	-2.14	0.0427
4	0.989	0.978	85.08	-0.81	0.4287
1m	0.998	0.996	78.12	0.48	0.6351
4m	0.996	0.991	76.03	1.54	0.1380
Upper First Molar Mesiobuccal Cusp y coordinate Alpha=0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.996	0.992	69.63	1.35	0.1889
2	0.995	0.989	68.88	-0.35	0.7285
3	0.997	0.993	69.36	-1.25	0.2235
4	0.994	0.987	67.46	-0.63	0.5318
1m	0.998	0.996	65.44	-0.70	0.4883
4m	0.996	0.993	72.51	0.01	0.9960

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

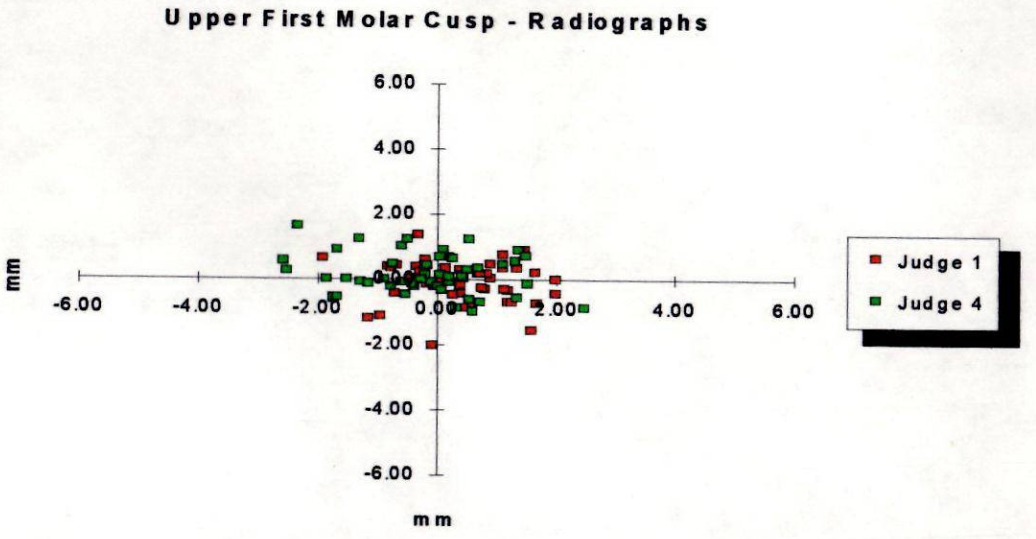


Figure 63. Error associated with the location of Upper First Molar Mesiobuccal Cusp for Judge 1 and 4 from the radiographs.

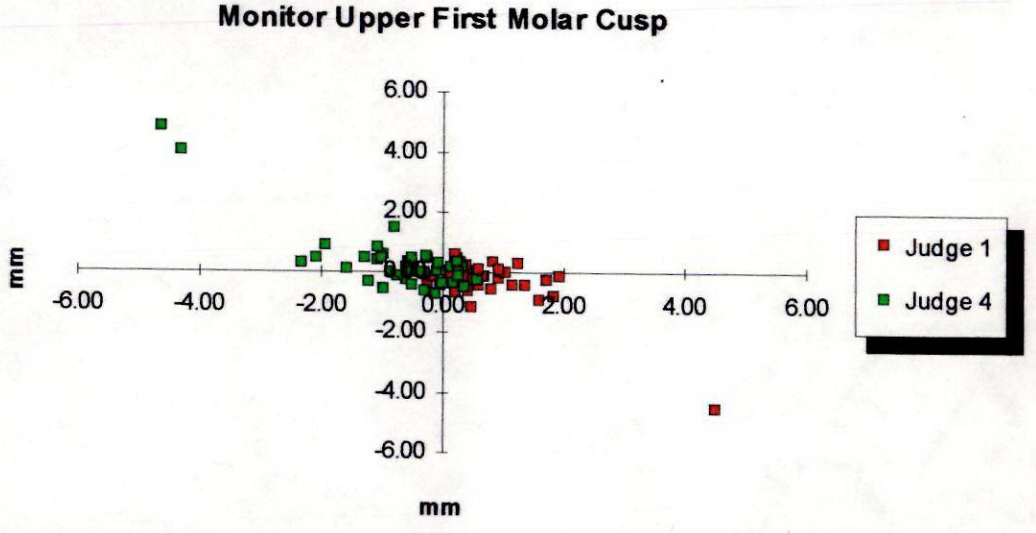


Figure 64. Error associated with the location of Upper First Molar Mesiobuccal Cusp for Judge 1 and 4 from the computer monitor.

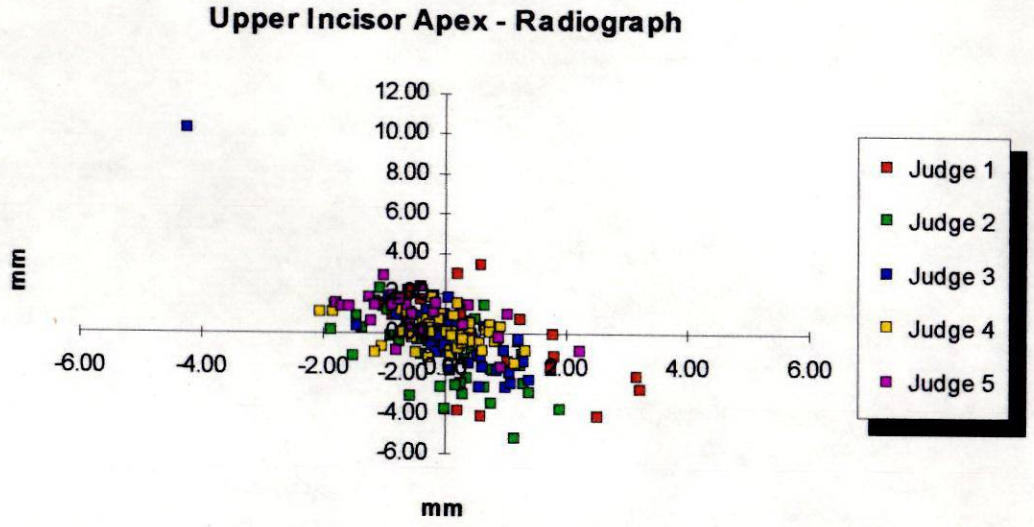


Figure 65. Error associated with the location of Upper Incisor Apex from all twenty five radiographs

Table 25.-- Intra-Examiner Reliability Upper Incisor Apex

Upper Incisor Apex x coordinate Alpha=0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.996	0.992	75.38	0.34	0.7365
2	0.991	0.982	78.68	1.79	0.0857
3	0.989	0.979	76.25	0.94	0.3569
4	0.997	0.994	77.20	-0.94	0.3567
1m	0.994	0.988	94.81	0.58	0.5647
4m	0.997	0.995	80.47	-0.94	0.3555
Upper Incisor Apex y coordinate Alpha=0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.992	0.984	86.25	-0.33	0.7456
2	0.971	0.942	78.72	-2.55	0.0177
3	0.953	0.908	77.00	-1.10	0.2825
4	0.996	0.992	84.91	-0.63	0.5368
1m	0.989	0.978	90.70	-0.78	0.4459
4m	0.996	0.991	87.60	-1.11	0.2811

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

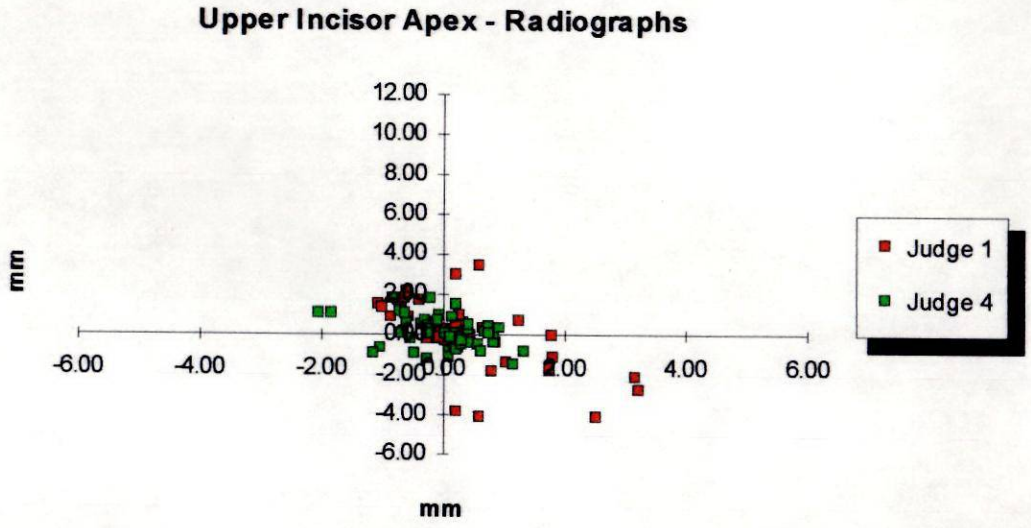


Figure 66. Error associated with the location of Upper Incisor Apex for Judge 1 and 4 from the radiographs.

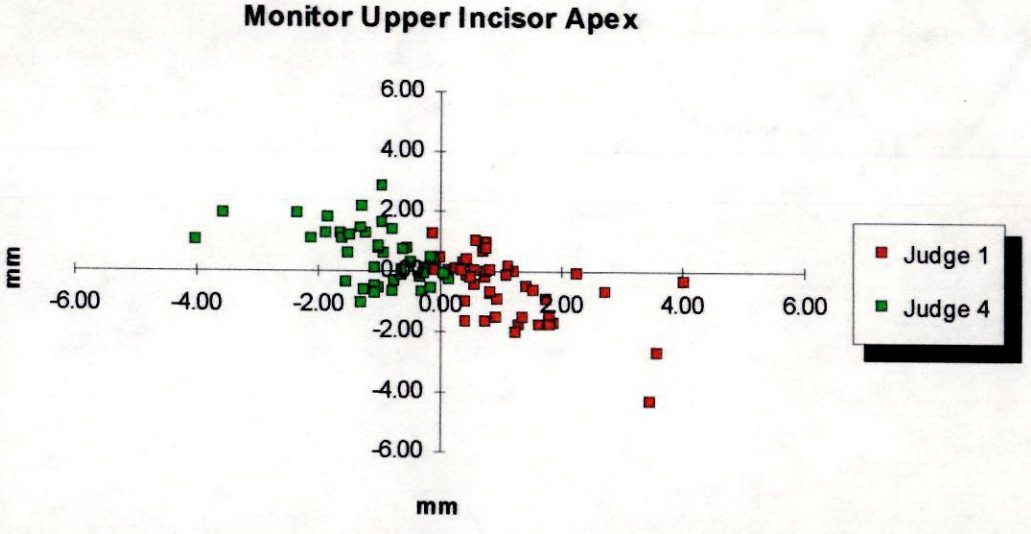


Figure 67. Error associated with the location of Upper Incisor Apex for Judge 1 and 4 from the computer monitor.

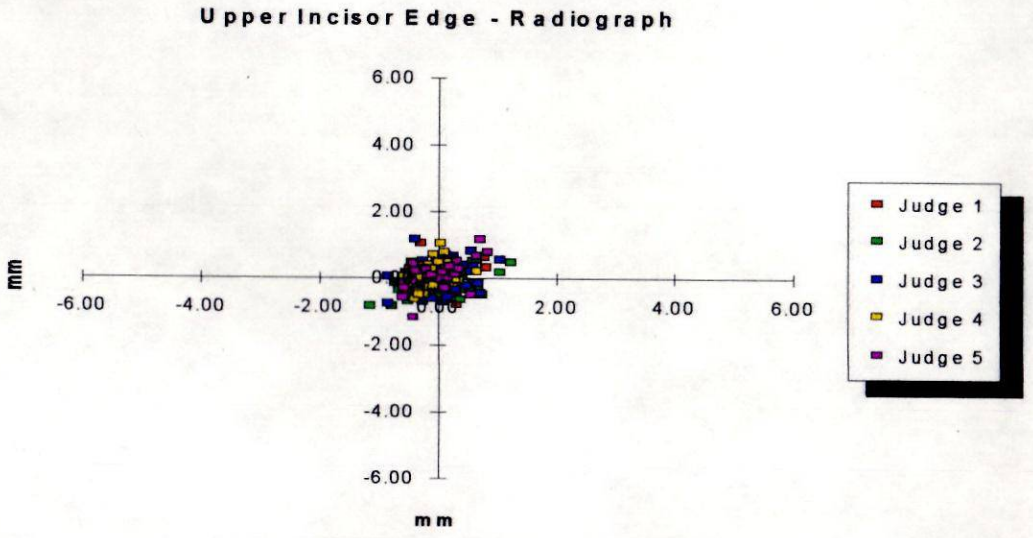


Figure 68. Error associated with the location of Upper Incisor Edge from all twenty five radiographs.

Table 26.-- Intra-Examiner Reliability Upper Incisor Edge

Upper Incisor Edge x coordinate Alpha=0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.998	0.996	76.92	1.86	0.0759
2	0.998	0.995	94.75	1.04	0.3072
3	0.998	0.996	94.84	-0.64	0.5275
4	0.999	0.999	95.34	-0.90	0.3796
1m	1.000	0.999	100.20	-0.38	0.7093
4m	0.999	0.999	100.12	1.18	0.2504
Upper Incisor Edge y coordinate Alpha=0.05					
Judge	r	r^2	Pooled Variance	t	P(T<=t) two-tail
1	0.996	0.992	75.3796	0.3404	0.7365
2	0.999	0.997	77.5779	0.1585	0.8754
3	0.998	0.995	76.0948	1.1606	0.2577
4	0.999	0.997	77.1615	-0.3218	0.7504
1m	1.000	1.000	76.8109	1.4676	0.1564
4m	0.999	0.999	78.5680	1.6348	0.1163

All measurements relate intra judge observations from first recording to second recording.
Judge 1,2,3 and 4 refer to measurements from original radiographs.
Judge 1m and 4m refer to measurements for Judges 1 and 4 from the monitor.
Correlation Coefficient (r). Reliability Coefficient (r^2)
P value for two tailed Paired two sample for means t test.

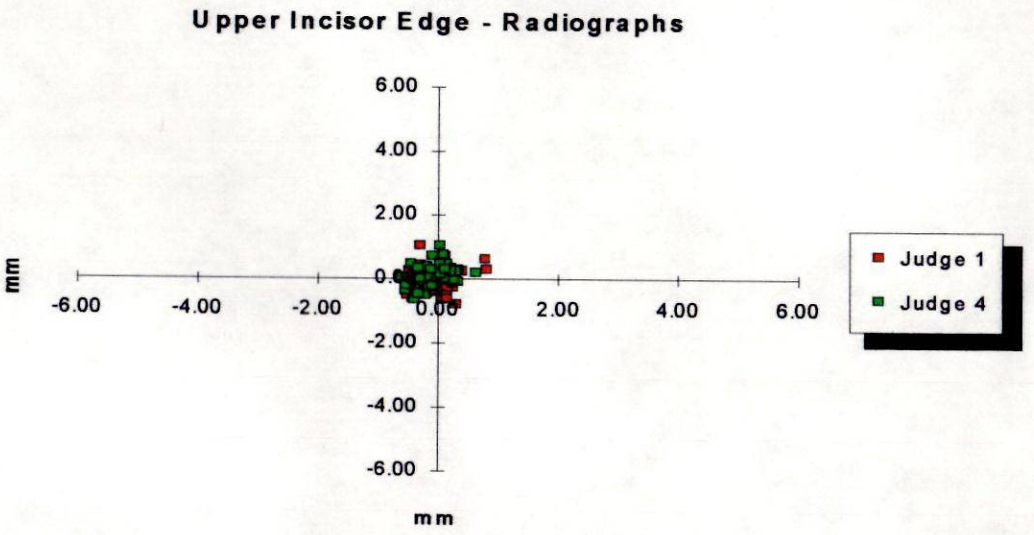


Figure 69. Error associated with the location of Upper Incisor Edge for Judge 1 and 4 from the radiographs.

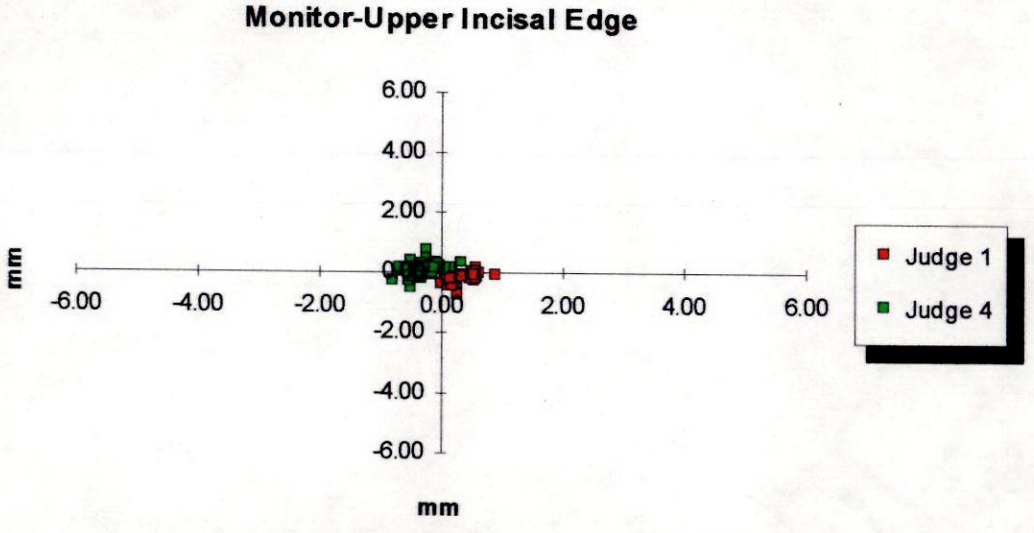


Figure 70. Error associated with the location of Upper Incisor Edge for Judge 1 and 4 from the computer monitor.

Table 27.-- Standard deviation for deviations from the mean of all original radiographs and electronic media (mm).

Landmark	Mean (x)	Std Dev (x)	Mean (y)	Std Dev (y)
A Point	0 00	0 69	0.00	1 28
A Point - Monitor	0.00	0 65	0.00	0.80
ANS	0 00	0.86	0.00	0 43
ANS - Monitor	0 00	0 72	0 00	0.29
B Point	0.00	0 32	0.00	1.12
B Point - Monitor	0 00	0.49	0 00	0 85
Condyle	0 00	1 75	0 00	4.41
Condyle - Monitor	0.00	0.43	0 00	1 13
Fiducial 1	0 00	0 19	0 00	0 26
Fiducial 1 - Monitor	0.00	0 42	0 00	0.31
Fiducial 2	0 00	0 18	0 00	0 23
Fiducial 2 - Monitor	0 00	0 36	0 00	0.94
Fiducial 4	0 00	0.00	0 00	0 17
Fiducial 4 - Monitor	0 00	0 46	0 00	0 29
Gonial Angle	0.00	1 11	0.00	1.45
Gonial Angle - Monitor	0 00	0.90	0 00	1 16
Lower First Molar Apex	0.00	1.16	0 00	1 09
Lower First Molar Apex - Monitor	0 00	1 23	0 00	1 01
Lower First Molar Cusp	0 00	0 92	0 00	0 69
Lower First Molar Cusp - Monitor	0 00	1 44	0 00	0 78
Lower Incisor Apex	0 00	1 23	0.00	1.33
Lower Incisor Apex - Monitor	0.00	1 44	0.00	2 56
Lower Incisal Edge	0 00	0.37	0.00	0 54
Lower Incisal Edge - Monitor	0 00	0 41	0.00	0 29
Menton	0.00	1 39	0 00	0 61
Menton - Monitor	0 00	0.76	0 00	1.25
Nasion	0 00	0.52	0 00	0 78
Nasion - Monitor	0 00	0.95	0 00	0 89
Orbitale	0.00	1 23	0 00	1.06
Orbitale - Monitor	0.00	1 13	0.00	0 76
Posterior Nasal Spine	0 00	3.29	0 00	0 66
Posterior Nasal Spine - Monitor	0 00	1 07	0.00	0 60
Pogonion	0.00	0 34	0 00	0 87
Pogonion - Monitor	0 00	0 45	0 00	1 68
Porion	0.00	1 23	0.00	1 79
Porion - Monitor	N/A	N/A	N/A	N/A
Sella	0 00	0 33	0 00	0.42
Sella - Monitor	0 00	0.21	0 00	0 28
Upper First Molar Apex	0 00	1 15	0 00	1.08
Upper First Molar Apex - Monitor	0 00	0 70	0 00	0 78
Upper First Molar Cusp	0 00	1 07	0.00	0 66
Upper First Molar Cusp - Monitor	0 00	1.25	0 00	1.02
Upper Incisor Apex	0 00	0 88	0.00	1 56
Upper Incisor Apex - Monitor	0 00	1.38	0 00	1 12
Upper Incisal Edge	0 00	0 36	0 00	0 38
Upper Incisal Edge - Monitor	0 00	0 37	0 00	0 22

DISCUSSION

Subjective Spatial Resolution Test - Monitor Bias

This test examined identical digital radiographic images shown simultaneously on Monitor A and Monitor B. There were a total of 10 tests of monitor bias for each judge, making a total of 280 responses. The majority of judges (55.36% or 155/280) chose the correct response (no difference). Some judges (15% or 42/280) chose Monitor A as being superior. Out of the 42 total responses selecting Monitor A, six judges made up 32 of these responses On closer examination, these same six judges chose Monitor B 16 times. This means these six judges between them only selected no difference between the monitors 2 times (out of a possible total of 60). This indicates these judges were uncomfortable in selecting no difference as a response. Of the judges who chose Monitor B as offering the image with the most information, 36 out of 83 responses came from six judges These results indicate that six judges preferred Monitor B, while six other judges seemed reluctant to choose 'no difference' between the monitors. Overall, the results were normally distributed which indicated there was no monitor bias.

Although the monitors used in this test were the same model, and manufactured in the same month, there was a perceptible difference between the monitors. This was overcome to some extent by using the monitor matching device previously described. Matching monitors exactly is extremely difficult. This is due largely to the enormous number of electronic components of a monitor.

The tolerances of each component capacitor and resistor is critical. Even a slight difference in tolerance will result in different visual output of the monitor (Rabne 1993). Differences in the phosphorous coatings on the screens will result in monitors that produce unequal display characteristics. Adding to the variability in display is the video graphics board which drives the monitor. Slight differences in even identical boards will result in monitor displays that are not identical. Currently there is not a simple solution available to match monitors exactly. In future years there is a possibility of computer graphic user interfaces (GUI) offering a standard process to predictably calibrate the monitors, however this is not currently available (Rabne 1993). In this test the monitor termed Monitor A had a slight blue tinge. While Monitor B was best described as having a slight brown tinge. Although there was a slight difference between the monitor displays, this did not bias the results

Low Versus Medium Resolution

This test examined low resolution (384 pixels X 256 pixels) lateral cephalometric images against medium resolution (768 pixels X 512 pixels) images. The result of this test conclusively demonstrated that the judges could select the medium resolution image as offering more information than the low resolution image. There were 10 responses (out of a total of 140) from judges who chose the low resolution image as being superior to the medium resolution image. Two judges were responsible for 7 of these responses. Both of these judges were quite inexperienced with computers, and in examining computer screens. It is ironic that the majority of judges could clearly see the difference between the medium and low resolution images, yet two of these judges consistently chose the low resolution film as offering the most information Possible explanations for this are· 1 The judges incorrectly recorded

their observation; 2. The judges criteria for information available from a lateral cephalometric image are at odds with the majority of orthodontists

It is quite apparent when viewing a 384 pixels X 256 pixels image of a lateral cephalometric radiograph, that there is not enough spatial resolution. At this resolution, and displayed to fill the size of the screen, the image looks like a collection of gray squares. To enable viewing of all three resolution films at the same size, the low, medium, and high resolution films were viewed at a screen ratio of 4:1, 2:1, and 1·1 respectively When a film is viewed at a ratio greater than 1.1, the computer compensates by stretching pixels to fill the selected area. There is no more information in a film when this is done, it just appears larger Each anatomical edge in a lateral cephalometric film was seen as a series of steps This series of steps was actually individual pixels.

High Versus Medium Resolution

When judges where shown the high resolution (1536 pixels X 1024 pixels) image and the medium resolution (768 pixels X 512 pixels) image their responses were rather interesting. The medium resolution image was chosen as offering more information 107 times (a response rate of 38.21%). The judges chose no difference between the images for 132 (out of a total of 280) responses, i e. a response rate of 47 14%. While the high resolution image was only selected as offering more information 41 times (14 64%)

The high resolution (1536 pixels X 1024 pixels) image had twice the spatial resolution of the medium resolution (768 pixels X 512 pixels) image. Yet, 85 35% of the responses showed there to be no difference between the images, or the medium resolution image offered more information It is clear from this test that clinicians who are familiar with lateral cephalometric radiographs did not choose the higher resolution

film as offering more information than the medium resolution film Hence, the medium resolution film has enough spatial resolution for clinicians to view the anatomy of interest. This finding is in broad agreement with Jackson et al (1985) and Oliver (1991). Both of these investigators suggested that a 600-800 square pixel image was sufficient for cephalometric needs. However, this decision was based on the investigators own subjective opinion, without any testing to validate their statement. This spatial resolution also correlates with the conclusion of medical radiologists examining lung radiographs when one takes into account the difference in the physical film dimensions (Cocklin et. al. 1983).

The medium resolution image was frequently chosen as being the superior image. This may have been due to the medium image appearing with more contrast than the higher resolution film This may represent the importance of gray scale resolution in digital lateral cephalometric films. The medium resolution films when displayed on the screen were actually viewed at a ratio of 2:1 This process effectively doubles the size of a given pixel. Each pixel has one gray shade associated with it, the possible number of gray shades is dependent on the gray scale resolution of the image (e.g. 8 bit images have 256 possible gray shades). Because the pixel size of the medium resolution images were double that of the high resolution films the effective shade changes from one pixel to another pixel was amplified in the medium resolution films. This apparent accentuation of shade changes may have been responsible for the increased contrast in the medium resolution films. The actual number of gray shades is greater in the higher resolution film shown at a viewing ratio of 1·1.

A paradox is presented. On one hand the medium resolution film is regarded as being of equal or superior information value than the high resolution film Yet, the high resolution film has twice the spatial detail of the medium resolution film An

increase in spatial resolution allows one to discern objects that are close to one another as separate entities. On the surface this appears a most desirable trait for cephalometric appraisal where many anatomical points of interest overlap one another The low, medium and high resolution images used in this test were all 8 bit images (i.e. 256 gray shades). However, by viewing the medium resolution image at a ratio of 2.1, the apparent gray scale contrast of the image was enhanced, although in reality it was still an 8 bit image. It is apparent that spatial resolution and gray scale resolution jointly contribute to the overall information quality of an image The next logical step in the analysis of resolution requirements for lateral cephalometric radiographs would be to examine gray scale requirements for lateral cephalometric films.

Digital images require substantial electronic storage space Doubling the spatial resolution of an image, requires a four fold increase in electronic storage requirements. By increasing the size of already large images, image storage and file handling properties are made much more difficult. It is important to store only the information necessary to use a digital representation of a lateral cephalometric radiograph in the same way as clinicians now use silver halide cellulose films. However, in the near future High Definition Television (HDTV) will become a common media, with the likelihood of becoming a standard across many industries. Speculating, this would mean the cost of this high resolution would remain relatively low, and it would be supported by the majority of computer hardware and software manufacturers for many years. The spatial resolution of this medium is 1536 pixels X 1024 pixels, that is the same resolution used in this study (termed high resolution)

Even though this study found the 768 pixel X 512 pixel resolution images to be acceptable to clinicians, I would recommend using the higher resolution 1536 pixels X 1024 pixels images This reasoning is in part due to my belief that in order to

determine an optimum spatial resolution, an optimum gray scale resolution needs to be found. Also with the advent of HDTV, one can expect many advances in image storage, file processing and display characteristics associated with these large image files.

Objective Spatial Resolution Test

Having determined an optimum spatial resolution of 768 pixels X 512 pixels images in the subjective spatial resolution test, cephalometric landmark location was then tested at this spatial resolution. It must be remembered that the images used in this study were photographs of the original films digitized and placed onto Photo CD ROM. This method of converting an analog cephalometric radiographic film into a digital medium is useful to test spatial resolution, but produces inferior digital images compared to other digital processes such as scanning. The medium resolution (768 pixels X 512 pixels) images were chosen to test landmark location, as even though the future of digital images will likely be in the high resolution (1536 pixels X 1024 pixels) category, if landmark location proved to be accurate at the lower resolution (768 pixels X 512 pixels) compared to the original radiographs then landmark location at the higher resolution (1536 pixels X 1024 pixels) must by definition be even more accurate.

In examining all of these landmarks, there is no gold standard measurement. It is not known what the correct landmark identification is. Consequently, landmark locations were plotted as a deviation from the group mean In examining these results, one can graphically visualize the differences in landmark location between different examiners, different points, and between different media (original radiograph and digital computer image).

Fiducial Data

The measurement of the fiducial points theoretically removes any subjective judgment on the part of the investigator when locating a landmark Fiducial 1 was the upper right hand fiducial point. This point was furthest from the origin, and the greatest error was expected here. From the radiographs there was quite a tight group of observations (+/- 0 19 mm in x the axis, and +/- 0 26 mm in the y axis) All fell approximately within a 0 5 mm diameter circle. Perhaps this 0 5 mm diameter represents the greatest accuracy in locating a known point with no subjective observer judgments having to be made, using the conventional cephalometric tracing technique with a 0.3 mm lead pencil. However, to state this several assumptions had to be made:

1. Minimal Measurement Error

The measuring instrument was accurate This was verified by taking a known sized object and measuring it by the digitizer technique. This proved accurate by a conventional calibrated ruler All measurements were double determined If an error greater than 0.3 mm occurred, these measurements were again measured twice, and this process repeated until both measurements were within the 0 3 mm range

2. Uniform Fiducial Pin Hole Size

Several of the fiducial pin holes were not of uniform size, slight variations were present. This may have been the result of puncturing the pinholes at different amounts of pressure, leading to puncture marks of slightly varying size The amount and direction of this error is small, but nonetheless present.

3. Radiographs completely flat when fiducials placed

Every endeavor was made to maintain the orientation of the film completely flat at the time of placement of the fiducials.

4. Permanent deformation of silver halide cellulose film

It is theoretically possible for some permanent or temporary deformation of the lateral cephalometric films to have occurred over time or with changing atmospheric conditions (Mensforth 1993) If this occurred it would result in a change of physical dimension from one radiograph to another

However, 25 lateral cephalometric radiographs were examined by five judges, four of these judges examined the films twice Effectively there were 350 estimations (175 x and 175 y coordinates) for each of the 25 points examined Having this enormous number of estimations significantly reduced the possible random error associated with any of the previously discussed sources of random error. Consequently, this data demonstrates any point clearly marked with no ambiguity, can only be reliably selected from a conventional cephalometric film within a 0.5 mm envelope of error. I believe this is the first time a report has been made of the accuracy of locating an unambiguous point on a cephalometric film. Baumrind et. al. (1971) initially described the use of fiducial landmarks In a later paper, Baumrind (1980) states that these fiducial points are also used to prevent errors of landmark location greater than 1mm occurring on multiple tracings of the same radiograph, this technique is thoroughly discussed in his paper (Baumrind 1980). However he did not report the actual accuracy of locating a fiducial landmark

The monitor fiducial data was not as closely packed as the original radiographs. However, part of this may be explained by how the landmarks were selected on the monitor At the spatial resolution of 768 pixels X 512 pixels, the smallest increment the mouse could be moved was 0.04 units This equated to a linear distance of approximately 0.45 mm. This meant that one could not identify a point any closer than 0.45 mm accuracy. However, even though the monitor fiducial data was

not as accurate as that from the radiographs, actual cephalometric anatomical point location exhibited a similar error distribution Even though the subjective spatial resolution test showed that clinicians could not identify the higher resolution film (1536 pixels X 1024 pixels) from the medium resolution film (768 pixels X 512 pixels), by moving to the high resolution film to choose landmarks, the accuracy of the mouse would be double that of the medium resolution film. The mouse could be moved a minimum of 0.25 mm accuracy This degree of accuracy is marginally superior to that of the conventional cephalometric radiographic technique which is limited to 0.3 mm (the width of a lead tracing pencil). Also, it is possible on a computer to zoom in on an area. One can even zoom beyond a 1:1 display of the actual spatial resolution of a landmark to enable a judge to choose a landmark much more easily

Anatomical Data

The error pattern associated with the location of the majority of landmarks from the original radiographs was similar to that reported by Baumrind (1971, 1980), and Broch et. al. (1981). These data support Baumrind's notion that each landmark seems to have associated with it a characteristic pattern of error For example in locating PNS, judges err more in the horizontal plane than in the vertical plane. The shape of the error is predictable, in that an observer may follow the palatal plane posteriorly until a position is reached which one feels is the PNS.

The size of the error in locating landmarks from a 768 pixel X 512 pixel image was approximately the same as when these same points are located on the original radiograph Some points have less error associated with their location on the monitor, for example Sella, PNS, Upper First Molar Mesiobuccal root apex, Orbitale and A

Point Other points were more accurately located on the original radiographs, for example Menton, Nasion, Pogonion and the upper incisor apex.

The direction of the error from the monitor was often different to that of the original radiograph. For some points on the computer monitor the error seemed to be at approximately a 45 degree angle, while on the original film the direction of the error was in the vertical direction, for example B Point The lower incisor edge has a spherical error distribution from the original film, but on the monitor data it had a 45 degree orientation. This common distribution of the monitor data at an acute angle may represent: systematic error; the mechanics of locating a landmark from the computer screen; or a combination of them both. All the original radiographic data points were corrected to a specific axis. However, no rotation correction was possible on the computer monitor, and it is possible that this may constitute some systematic error that may be responsible for skewing the monitor data. The orthodontic profession has had over fifty years to develop and improve the lateral radiographic cephalometric technique. Anatomic landmark location from a computer monitor lacks a set of rules or conventions for systematic landmark location As orthodontists learn new skills in locating anatomic landmarks from computer monitors, the accuracy and reproducibility of their results will improve

CONCLUSIONS

Digital cephalometric radiographic images will replace silver halide cellulose radiographic films for routine use in orthodontics in future years. This will occur because it will reduce radiation exposure to a patient. In addition, storage, transport, and enhancement of digital media is easily accomplished. The orthodontic profession must define digital image resolution standards This must occur before the already flourishing commercial vendors have sold many of their own costly proprietary imaging solutions, which may or may not satisfy clinical diagnostic accuracy A standard needs to be set that is independent of any computer company's hardware or software, so a lateral cephalometric film can be sent digitally from one practitioner to another practitioner for clinical evaluation, insurance, case transfer or research.

This investigation clearly demonstrated that clinicians preferred digital cephalometric images with a spatial resolution of 768 pixels X 512 pixels. However, this may not be the optimum spatial resolution for analysis of lateral cephalometric radiographs From the objective test, landmark location was found to be essentially as accurate as using the original films. However, location of the fiducial landmarks was significantly less accurate with the electronic images on the monitor. Clinicians could not subjectively gain any more information from an image with double that spatial resolution. However, the gray scale resolution question remains unanswered With the advent of HDTV, 1536 pixels X 1024 pixels images seem currently to be the best choice But gray scale resolution requirements must be determined, and interpreted

along with the spatial resolution to find an image resolution suitable to satisfy clinical needs, and those of the researcher

APPENDIX

APPENDIX A

Landmarks used in the cephalometric study from Baumrind (1971) The following directions were given to all judges selecting landmarks from the various media.

Directions for cephalometric point location.

Securely fix the lateral cephalometric radiograph to a viewbox with four pieces of tape; one at each corner. Mount an ordinary piece of 0.003 matte tracing acetate upon the head film. Secure the tracing paper to the headfilm at each corner with additional tape. The desired set of anatomic landmarks (21) and the four pin holes in the headfilm are located and marked on the acetate with a sharp pencil in the following prescribed order A 0 3 mm pencil should be used and its point must be kept sharp.

4 fiducial reference and 21 landmarks marks are chosen in the following order:

Fiducial Reference Marks

1. Fiducial 1

This is the pin hole in the radiograph located at the right superior corner of the film

2. Fiducial 2

This is the pin hole in the radiograph located at the right inferior corner of the film.

3. Fiducial 3

This is the pin hole in the radiograph located at the left inferior corner of the film.

4. Fiducial 4

This is the pin hole in the radiograph located at the left superior corner of the film

Anatomic Landmarks

1. Sella

Sella Turcica, the midpoint of the pituitary fossa as determined by inspection

2. N

Nasion, the most antero-inferior point on the frontal bone at the nasofrontal suture.

3. Or

Orbitale, the most inferior point of the orbital margin, using the more prominent of the orbital images

4. ANS

Anterior Nasal Spine, the anterior tip of the sharp bony process of the maxilla at the lower margin of the anterior nasal opening.

5. PNS

Posterior Nasal Spine, the posterior spine of the palatine bone containing the hard palate.

6. UI edge

Upper central incisor edge, the tip of the incisal edge of the more anteriorly placed upper central incisor.

7. UI apex

Upper central incisor apex, the point of intersection between the long axis of the more anteriorly positioned upper central incisor and the contour of the tooth's root and curvature.

8. LI edge

Lower central incisor edge, the tip of the incisal edge of the more anteriorly placed lower central incisor.

9. LI apex

Lower central incisor apex, the point of intersection between the long axis of the most anteriorly positioned lower incisor and the contour of the tooth's root end curvature.

10. UFM cusp

Upper first molar mesiobuccal cusp, (the average value between the left and right), the most occlusal point of the mesiobuccal cusp of the upper first molar.

11. UFM apex

Upper first molar mesiobuccal root apex, (the average value between the left and the right), the most apical point on the mesiobuccal root of the upper first molar

12. LM cusp

Lower first molar mesiobuccal cusp (the average value between the left and the right) of the most occlusal point mesiobuccal cusp of the lower first molars Note the more mesially positioned will usually be the right, and the more distally positioned will usually be the left

13. LM apex

Lower first molar mesiobuccal root apex, (the average value between the left and the right), the most apical point of the mesiobuccal root of the lower first molar

14. A

Point A, the deepest point on the curvature of the surface of the surface of the maxillary bone between ANS (Anterior Nasal Spine) and the alveolar crest of the upper central incisor.

15. B

Point B, the deepest point on the curvature of the anterior border of the mandible between Pogonion and the alveolar crest of the lower central incisor.

16. Pg

Pogonion, the most anterior point on the bony chin at the midline

17. Upper Go

Upper Gonion, or Gonion for the left side. This is a point on the curvature of the angle of the mandible located by bisecting the angle by lines tangent to the posterior ramus and the inferior border of the mandible. The lowest point on the curvature of the angle of the mandible where the body of the mandible meets the ramus. Estimates were made for both sides of the mandible. The more superior value is considered to represent the left side

18. Lower Go

Lower Gonion, or Gonion for the right side The lowest point on the curvature of the angle of the mandible where the body of the mandible meets the ramus. Estimates were made for both sides of the mandible The more inferior value is considered to represent the right side.

19. Mechanical porion

Found by using Broadbent template Place template over ear rods, and head support, then mark the point representing mechanical Porion.

20. Condyle

Condyle point, the perceived central point of the mandibular condyle, averaged between left and right sides.

21. Me

Menton, the most inferior point on the mandible at the symphysis

Appendix B

Response sheet used for the subjective spatial resolution test

Bolton Brush

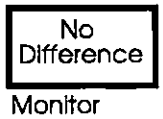
Growth Study Workshop

Dear Doctor,
Thank you for your participation in this study.

Name: .....

Instructions.

Thirty image pairs will be displayed on the computer monitors.
Please examine the radiographs displayed on monitors A and B, and choose only **one** of the following three alternatives.

Please check: 

 if the images are of equivalent quality, or

please check: Monitor 

 if the image on monitor A is superior, or

please check: Monitor 

 if the image on monitor B is superior.

Image Pairs:

#1 

Monitor 


Monitor 



#2 

Monitor 


Monitor 



#3 

Monitor 


Monitor 



#4	No Difference ☐	Monitor  ☐	Monitor  ☐
#5	No Difference ☐	Monitor  ☐	Monitor  ☐
#6	No Difference ☐	Monitor  ☐	Monitor  ☐
#7	No Difference ☐	Monitor  ☐	Monitor  ☐
#8	No Difference ☐	Monitor  ☐	Monitor  ☐
#9	No Difference ☐	Monitor  ☐	Monitor  ☐
#10	No Difference ☐	Monitor  ☐	Monitor  ☐
#11	No Difference ☐	Monitor  ☐	Monitor  ☐
#12	No Difference ☐	Monitor  ☐	Monitor  ☐
#13	No Difference ☐	Monitor  ☐	Monitor  ☐
#14	No Difference ☐	Monitor  ☐	Monitor  ☐
#15	No Difference ☐	Monitor  ☐	Monitor  ☐

#16	No Difference ☐	Monitor  ☐	Monitor  ☐
#17	No Difference ☐	Monitor  ☐	Monitor  ☐
#18	No Difference ☐	Monitor  ☐	Monitor  ☐
#19	No Difference ☐	Monitor  ☐	Monitor  ☐
#20	No Difference ☐	Monitor  ☐	Monitor  ☐
#21	No Difference ☐	Monitor  ☐	Monitor  ☐
#22	No Difference ☐	Monitor  ☐	Monitor  ☐
#23	No Difference ☐	Monitor  ☐	Monitor  ☐
#24	No Difference ☐	Monitor  ☐	Monitor  ☐
#25	No Difference ☐	Monitor  ☐	Monitor  ☐

Thank you for your time, effort and consideration for research.

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