T-BME/38/4//44245

Real-Time Sensing of Tooth Position for Dental Digital Subtraction Radiography

Grigore C. Burdea
Stanley M. Dunn
Charles H. Immendorf
Madhumita Mallik

Reprinted from
IEEE TRANSACTIONS ON BIOMEDICAL ENGINEERING
Vol. 38, No. 4, April 1991

Real-Time Sensing of Tooth Position for Dental Digital Subtraction Radiography

Grigore C. Burdea, *Senior Member, IEEE*, Stanley M. Dunn, *Member, IEEE*, Charles H. Immendorf, and Madhumita Mallik

Abstract—**Dental digital substraction radiography requires accurate repositioning of the patient and X-ray source in order to facilitate correct diagnostic of bone loss. Present mechanical repositioning systems do not allow radiography of posterior teeth, and are uncomfortable for the patient. A new repositioning system that utilizes a six degrees of freedom position sensor and a robot arm with X-ray source is proposed. A mathematical model for the system is given, and the robot arm solution is obtained based on patient position. An error analysis is performed in order to determine the influence of sensor and robot errors on system accuracy. A series of experiments to determine sensor noise and accuracy are described. These tests showed relatively small errors over the work envelope of the sensor. Further tests showed that there is no adverse effect due to the presence of metal work in the patient's mouth. The high bandwidth of the sensor may allow real time tracking of small movements of the patient's head.**

I. Introduction

DENTAL radiographs are one of the most widely used diagnostic tools in dentistry. This remains true in spite of the inherent limitation that they provide only a two-dimensional projection (i.e., view) of the area of interest. For many disease processes this two-dimensional view is sufficient to characterize the pathology and initiate a treatment plan. For detecting the presence or absence of lesions, one need only look for intensity changes in the film, or image if it has been digitized.

However, as dental health has improved, the emphasis has shifted toward early detection of disease, requiring more exacting instruments. Nowhere is this more evident than in the diagnosis and treatment of periodontal disease. The goal is to be able to detect as early as possible, small changes in the bony structure supporting the tooth. If it is not caught in time, then the result can be tooth loss and continued oral health problems.

Studies have shown that disease can be detected earlier and with greater accuracy by looking at the difference between radiographs taken over time, instead of a single radiograph. This "subtraction" technique can be done optically by aligning films or digitally, by digitizing each film and then subtracting corresponding picture elements to compute a difference image. With the advent of personal computers and low cost imaging systems, most work can be done digitally.

This presents a unique imaging problem. The two radiographs must enclose the same field of view of the mouth and must be taken with the same geometry. If they are not, then the radiographs cannot be subtracted meaningfully. The purpose of this subtraction is to eliminate the anatomic features not of interest, the so-called *structured noise* in the image. If the imaging geometry is not the same then the visual appearance of the structured noise is different and the difference will not be zero, yielding a false positive difference.

The goal must be to produce standardized views of the area of interest. Much work to date has been done on mechanical devices to fix the object (i.e., patient) and imaging device (X-ray source) in position, a description of relevant work being given in Section II. The goal of our research is to continue to study an approach to this problem that does not require fixing the patient in position relative to the X-ray source. The proposed approach is based on accurately positioning the X-ray source and then recognizing three-dimensional properties of the objects of interest that are invariant in any view of the object. Once these invariants are recognized and measured in a digital image, the two digital images can be aligned to produce standardized digital images (radiographs). Then an accurate subtraction can be made to remove structured noise.

The impact of this result goes beyond the application on radiology. Rather, this application motivated the study of the imaging problem and provides a wealth of data which can be used in experiments to quantify the accuracy, resolution and reproducibility of the approach. The same techniques are applicable in any other application with any source of image data. The general scientific problem being studied is one of extracting invariant three-dimensional object characteristics and using them impact dentistry, other applications of radiology and general imaging sciences.

Section II gives some background on tomographic image formation using X-rays. Section III is a description of the mathematics of measuring 3-D position, while results of preliminary studies showing the accuracy, reproducibility, and resolution of 3-D measurements are shown in Section IV. Concluding remarks are given in Section V.

II. Background and Significance

Digital subtraction radiography is one example of an imaging application requiring a differential diagnosis—a measurement of change over time. Two images, each of which records the state of the dentition at a specific time are taken. Since the normal anatomical structure should not change between the two films, the observed differences are indicative of growth or decay in the bony structure. The early studies [27], [25] used subtraction to detect lesion in between teeth and in the supporting structure of the teeth. The more recent papers [3]–[6] cite applications of digital subtraction to measure bone loss and density in the supporting structure of the teeth. Showing the density changes in color [4] improved the diagnosis agreement among observers. In all cases, however, the application papers point out the difficulty in making the initial measurements, the difficulty in reproducing the original imaging geometry and why this limits

Manuscript received May 3, 1990; revised August 16, 1990. This work was supported by grants from the University of Medicine and Dentistry of New Jersey and the Rutgers University Research Council.

G. C. Burdea, C. Immendorf, and M. Mallik are with the College of Engineering, Rutgers University, Piscataway, NJ 08855

S. M. Dunn is with the College of Engineering, Rutgers University, Piscataway, NJ 08855 and the University of Medicine and Dentistry of New Jersey, Newark, NJ 07107.

IEEE Log Number 9144245

0018-9294/91/0400-0366$01.00 © 1991 IEEE

the use of digital subtraction radiography. The same, or similar problems, arise in spatially disparate instead of time disparate image sequences. The correspondence problem is difficult to solve and the data acquisition should be examined first.

One of the earliest papers in the area by Webber et al. [32] described the principle that the subtraction process removed "noise" that is the unchanged anatomy during the time between the two films. Webber and his colleagues already recognized the requirements of fixing the geometry to produce standardized radiographs, and used a template to fix the mouth in position. Mechanical devices to standardize the imaging geometry were used by Grondahl [14], [15], Hausmann and colleagues [24], and Janssen [19], [20]. The studies by Grondahl and Hausmann and Janssen all used subtraction in trials to quantitate bone loss. Janssen [19] found that the digital subtraction system was the most sensitive to measure subtle change when compared to using a single radiograph or photographic subtraction, but required standardized geometry.

The application problem of standardizing geometry for digital subtraction radiography leads us to study the engineering problem of standardizing imaging geometry and calibrating the geometry. The problems of correspondence and registration have long been studied and widely reported in the computer vision field in, for example, the books [2], [18], [23], [28] and the papers [11], [1], [21], [29], [26].

The study of radiographs in not new. Some of the earliest work in image understanding and interpretation was done by Prewitt and Sklansky on radiographs. However, little attention was paid to the image formation geometry at that time (or even until now) because the applications had not warranted it.

In the original subtraction paper by Webber [32] the authors noted that there were four sources of error leading to improper registration of the pair of radiographs: tissue changes, film, X-ray energy, and inexact replication of imaging geometry. The first cannot be controlled other than by assumptions on the localization of the changes. The second can be controlled by the film type and processing chemicals. Webber later studied the effects of X-ray energy in [33] and showed that the effects of energy can be controlled. The remaining problem is that of controlling the imaging geometry.

To date, most approaches to this problem have relied on a model of radiograph formation of a point projection of X-rays along straight lines through the tissue. The X-rays are attenuated along these diverging straight line paths and form a distorted image on the film behind the hard tissue. As the source and/or film move with respect to the tissue, the appearance of the tissue on the film is changed nonlinearly. Thus, to generate radiographs with the same appearance it is important to reproduce the original imaging geometry. Following this line of reasoning, many studies of the subtraction technique have been reported that use mechanical fixtures [32], [16], [24], [22], [20] and/or compensatory algorithms [22], [30] to control the imaging geometry, i.e., the position of the point source and the attenuation paths through the tissue.

A good example of a mechanical standardization system for subtraction radiography is given in [22]. The X-ray source, patient, and film are connected using an occlusal stent and cephalostat. The mechanical connection of the source, object, and film should restrict any variations to be in plane translations and rotations. By matching three anatomical features in the two films, the translation and rotations can be eliminated. The difficulty is twofold: first, the use of a mechanical fixture or fixed imaging geometry restricts the field of view and therefore only a limited portion of the dentition can be imaged. Second, disease processes are three and not two-dimensional.

Digital subtraction is dependent on our ability to precisely match imaging geometry. Instead, if one can gather enough information to create a 3-D model of the tissue then 2-D projections can be synthesized; subsequent radiographs with unknown imaging geometry can be matched.

One approach to creating three-dimensional models recently on the dental literature is tomosynthesis [31]. A finite set of radiographs are taken with the source at known locations on a circular locus in a plane on one side of an object. A reconstruction from projections is not explicitly done, rather the projection data is used to synthesize an arbitrary projection. Arbitrary projections are linear combinations of the projections taken [31]. It is easy to see that tomosynthesis is a nice extension of axial transverse tomography, but instead of rotating the patent and film as described in Herman [17], the source is moved in a circle on the one side of the object. The advantages of this method are that any projection can be synthesized and thus this technique can be used to generate the projection needed for a subtraction and that three-dimensional data is collected and used. The disadvantage is the number of images that have to be taken and the time required for the process. Currently, tomosynthesis is used only in the laboratory on dry skulls to demonstrate the feasibility of the approach [30]. To use this in practice would required a CCD array or a film pack to gather the set of images. It would, of course, be difficult to ensure that a patient could be fixed in position in the time period between changing single films, rendering tomosynthesis useless if one has to take a set of individual films.

The advantage is that the tomosynthetic three-dimensional model can be created once and compared to a radiograph taken at a later time from an arbitrary projection. In this way the imaging geometry of the second radiograph need not be exactly that of the tomosynthesis images. The cost is the precision with which the tomosynthesis projections have to be taken. Using current technology this means the patient must remain fixed in position between each projection image. Furthermore, the entire set of tomosynthesis images must be stored for later use.

The tomosynthetic paradigm of synthesizing 2-D projections from a 3-D model still relies on the point projection model of radiograph formation. This model describes how the appearance of the tissue changes in the 2-D radiograph as a function of the 3-D position of the tissue. In essence, this approach to subtraction radiography relies on reproducing a description of how tissue changes appearance in a radiograph. Mechanical control of the geometry is promising, but limits the routine use of subtraction radiograph. There is an alternative model that may not require such rigid mechanical standardization in practice. Instead of modeling radiograph formation by how things change appearance, model the formation by describing what geometric (and intensity or attenuation) relations remain the same. Thus, even though there may be nonlinear distortion between a pair of radiographs, there are relationships that remain the same in both films. Modeling the imaging geometry by what remains the same, or *invariant*, has some advantages.

The imaging geometry model described in [12] uses projective invariants of points on a plane in 3-D. These are relations amongst points in the plane that remain the same regardless of how the plane may be positioned or oriented with respect to the X-ray source. In [12] a set of 88 periapical films with between 0–32° angulation error (about each of the three axes) and between 0–16 mm translation were used. Dunn and van der Stelt tween 0–16 mm translation were used. Dunn and van der Stelt showed that in each film the same invariant relationship can be measured between root apices and cemento-enamel junctions.

Standardized digital films can be produced by identifying the projective invariants in each of the two radiographs. The standardization can be done by finding the prespective projection that transforms the points of the invariant in one radiograph into the points of the (same) invariant in the second film. Thus, mechanical fixtures to register and align the films will no longer be needed and digital subtraction can be done in a clinician's office.

In order to do accurate densitometry work (such as computing lesion volume), the notion of invariants will have to be extended to differential invariants [34]. These are invariant relations of function values (i.e., image intensities, i.e., attenuation) of points in three dimensions. The differential invariants are a more precise description of the 3-D and 2-D relationships of positions of points and their image intensities.

The long term goal of developing a practical dental digital subtraction radiography system relies on an image formation model and a methodology for using it. The model of invariant relations described in [12] requires only the tissue of interest appear wholly contained in each of the two films. This is a much weaker constraint than requiring that the imaging geometry be reproduced exactly and can be achieved with simple control of existing radiograph systems. If the location of the patient and the desired imaging geometry are known, then a control system can reposition the X-ray source. This control system to reposition the source need only be accurate enough to guarantee that the tissue of interest will appear on the film. In this paper we shall describe the design of such a position sensing and control system and report on the accuracy of sensing 3-D position of the mouth.

III. Proposed Robotic System for Dental Digital Subtraction Radiography

Digital subtraction radiography requires a system that provides good repeatability and accuracy for patient positioning. Present dental subtraction radiography positioning systems are mechanical, uncomfortable for the patient, and impractical for certain tooth positions. We propose to replace present mechanical systems with a sensorized system without direct mechanical link with the X-ray source. This new system utilizes a robot arm as positioning device that tracks a magnetic sensor attached to the patient mouth.

Industrial robot systems are presently used in many engineering fields due to their programming flexibility, as well as excellent repeatability. An industrial robot can be viewed as a computer-controlled manipulator consisting of several rigid links connected in series by revolute or prismatic joints [13]. One end of the chain of manipulator segments is attached to a fixed supporting base, while the other end is free and equipped with a tool. Most robots have six degrees of freedom and their wrist can position and orient the attached tool. In order to adapt to changes in its environment the robot uses external sensing data given by vision, touch, force, or other types of sensors. By using this sensor data the robot can change strategies, grasp or release objects, or avoid colliding with obstacles.

The dental subtraction radiography requirement for conservation of patient X-ray source geometrical relationship led us to the idea to integrate an X-ray source with a robot manipulator. This proposed robotic dental subtraction radiographic system is shown in Figure 1. The weight of the X-ray source (about 20 kg) is within the payload capability of several existing manipulators. The source is mounted directly on the robot wrist, at the end of the manipulator. The symmetry of the X-ray beam can be exploited by aligning the robot wrist rotation axis (roll) with the lingitudinal axis of the X-ray source. Due to this symmetry the rotation angle of the wrist becomes irrelevant. A robot with only five degrees of freedom will suffice for this application, which reduces the cost of the system.

In dental radiography applications the position of the robot wrist with the attached X-ray source has to change as a function of the patient position. If a sensor can provide the position and orientation of the targeted tooth, then the robot can orient itself to follow that sensor and the tooth. The proposed system utilizes a six degrees of freedom sensor that provides position and orientation data (X, Y, Z, roll, pitch, and yaw) that is needed to position the robot. A high sensor bandwidth allows for small motions of the patient head to be compensated by the robot in real time.

In order to do real time tooth tracking, the amount of computation performed by the robot controller needs to be reduced. One way to reduce the amount of computation is to maintain a fixed geometrical relationship between tooth and sensor. Our system utilizes a mouth piece which serves as support for both the sensor and the X-ray film. Since the mouth piece is rigid, the geometrical relationship between sensor and the tooth is fixed, no matter how the patient turns his/her head. This relationship is computed off-line and stored in the memory of a host computer (such as a PC), which is also part of the system. The host computer downloads it to the robot controller, at the time of subsequent radiographs. While bone disease may alter over time (months) the geometry of the mouth, the presence of the mouth appliance assures a constant positioning of the X-ray source versus the film. Very large changes in mouth geometry will require of course a new mouth appliance, and the process described above in repeated.

A. Sensor Model

This section gives a model for the position sensing mechanism, as well as the robot target solution. This model is based on the homogeneous transformation technique first proposed by Denavit and Hartenberg [9]. This well known technique utilizes matrix algebra to represent the spatial geometry of the robot arm links with respect to a fixed reference coordinate system. If the original coordinate system OXYZ is rotated and translated in three dimensional space to form a new coordinate system OUVW, then the vector P can be transformed between the two coordinate systems through a homogeneous transformation matrix $\boldsymbol{T}$.

$$P_{xyz} = {}^{xyz}\boldsymbol{T}_{uvw} P_{uvw} \tag{1}$$

$\boldsymbol{T}$ can be partitioned into two submatrices namely, the rotation matrix expressing the orientation of the OUVW system versus OXYZ, and the position vector of the origin of OUVW in OXYZ. Its structure is expressed as

$${}^{xyz}\boldsymbol{T}_{uvw} = \begin{bmatrix} R_{3\times 3} & P_{3\times 1} \\ 0 & 1 \end{bmatrix} \tag{2}$$

where $P = [p_x, p_y, p_z]^T$. If a series of n rotations and translations are performed between multiple coordinate systems, then a composite homogeneous transformation matrix can be ob-

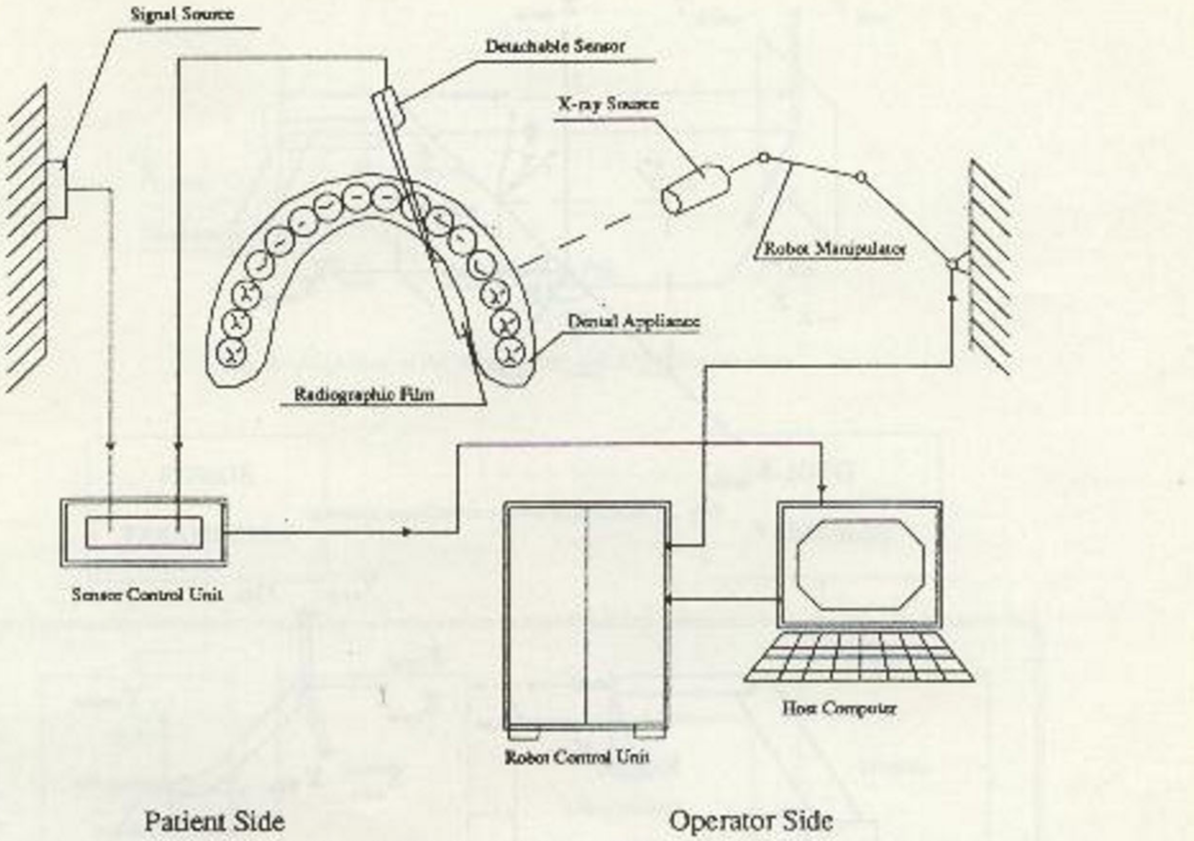

Fig. 1. Proposed robotic system for dental subtraction radiography.

tained by multiplying the sequence of homogeneous matrices ${}^{i-1}T_i$, with $i = 1, 2, \cdots, n$.

The sensor transmits six parameters. The first three are x, y, z representing the translation between sensor system of coordinates and its source system of coordinates, as shown in Fig. 2. The other three parameters are ψ, θ, and ϕ. ψ is the rotation of the x and y coordinates about the z-axis. θ is the rotation of the z and the rotated x coordinates about the rotated y-axis, and ϕ is the rotation of the rotated y and z coordinates about the rotated x-axis [10]. With these six sensor parameters, the 4×4 transformation matrix ${}^{source}T_{sensor}$ expressing the position and orientation of the sensor with respect to its source is given by (3) [10].

$$ {}^{source}T_{sensor} = T_{Z,\psi} T_{Y',\theta} T_{X'',\phi} T_{trans} $$

$$ = \begin{bmatrix} c\psi c\theta & c\psi s\theta s\phi - s\psi c\phi & c\psi s\theta c\phi + s\psi s\phi & xc\psi c\theta + y(c\psi s\theta s\phi - s\psi c\phi) + z(c\psi s\theta c\phi + s\psi s\phi) \\ s\psi c\theta & s\psi s\theta s\phi + c\psi c\phi & s\psi s\theta c\phi - c\psi s\phi & xs\psi c\theta + y(s\psi s\theta s\phi + c\psi c\phi) + z(s\psi s\theta c\phi - c\psi s\phi) \\ -s\theta & c\theta s\phi & c\theta c\phi & -xs\theta + yc\theta c\phi + zc\theta c\phi \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (3) $$

Another transformation ${}^{sensor}T_{tooth}$ expresses the position of the targeted tooth versus the sensor on the dental appliance. This transformation depends on the characteristics of the patient and therefore has to be determined for each patient, once the dental appliance is built. The transformation which gives the position of the tooth of interest in source coordinates is

$$ {}^{source}T_{tooth} = {}^{source}T_{sensor} \, {}^{sensor}T_{tooth} \quad (4) $$

While ${}^{sensor}T_{tooth}$ is fixed (but patient dependent), ${}^{source}T_{sensor}$ will vary as a function of head position and orientation. In this way, ${}^{source}T_{tooth}$ tracks the position of the patient's tooth, with respect to the fixed system of coordinates of the source.

B. Robot Model

Consider a robot manipulator with n degrees of freedom, as shown in Fig. 3. $n + 1$ orthonormal Cartesian coordinate systems are assigned to the robot links and base at the joint axes. The transformation ${}^{base}T_{wrist}$ that expresses the position and orientation of the robot wrist with respect to the robot base (commonly known as the "arm matrix") is

$$ {}^{base}T_{wrist} = {}^{base}T_1 \cdots {}^{n-1}T_{wrist} \quad (5) $$

Another transformation that is needed is ${}^{wrist}T_{x\text{-ray}}$ between the system of coordinates attached to the X-ray source and that attached to the robot wrist. In order to take advantage of the axial symmetry of the problem $OX_{x\text{-ray}}Y_{x\text{-ray}}Z_{x\text{-ray}}$ is chosen so that $Z_{x\text{-ray}}$ points towards the wrist, and coaxial with the wrist. The origin of $OX_{x\text{-ray}}Y_{x\text{-ray}}Z_{x\text{-ray}}$ is located at a fixed distance L from the robot wrist, corresponding to the distance to the pa-

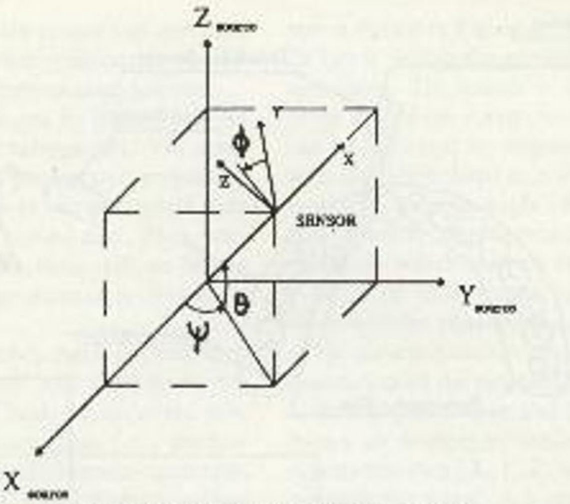

Fig. 2. Sensor and source coordinate systems.

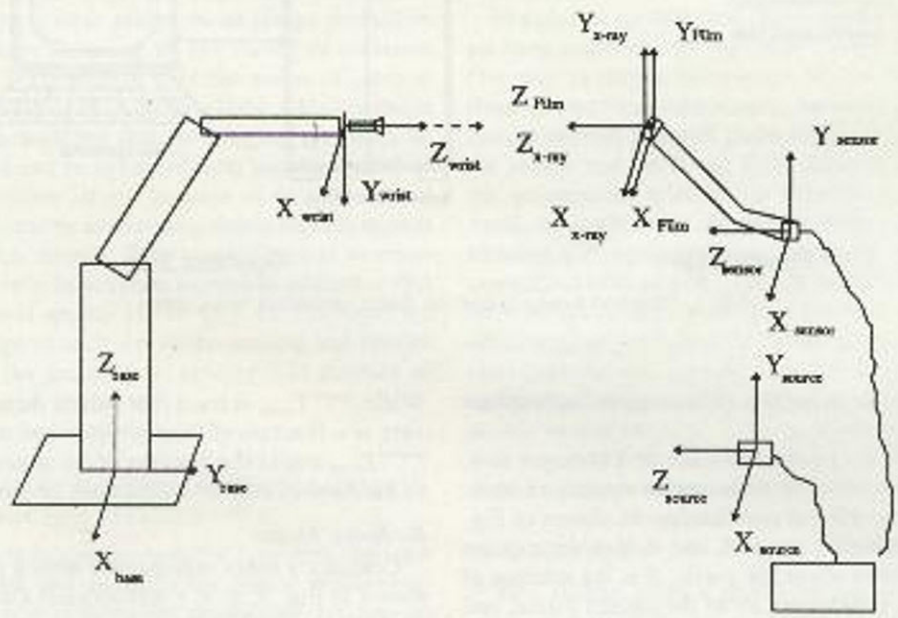

Fig. 3. Configuration of the robot manipulator.

tient. ${}^{wrist}T_{x\text{-ray}}$ is then fixed and given by

$$ {}^{wrist}T_{x\text{-ray}} = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & -1 & 0 & 0 \\ 0 & 0 & -1 & L \\ 0 & 0 & 0 & 1 \end{bmatrix}. \tag{6} $$

In order to orient itself the robot needs to know the patient position given in its own robot coordinates as ${}^{base}T_{tooth}$. If the sensor source is placed at a fixed and known location versus the robot base, then the transformation ${}^{base}T_{source}$ is also fixed and may be determined by measurements. The position of the patient with respect to the robot is then given by (8):

$$ {}^{base}T_{tooth} = {}^{base}T_{source}\ {}^{source}T_{tooth}. \tag{7} $$

The alignment between the X-ray source and the targeted tooth is one of the keys to the reproducibility of the images. One possible solution is for the X-ray source to be aligned perpendicular to the plane of the film, so that the two systems $OX_{x\text{-ray}}Y_{x\text{-ray}}Z_{x\text{-ray}}$ and $OX_{film}Y_{film}Z_{film}$ coincide. This is shown in more detail in Fig. 4. The alignment condition sufficient to obtain the robot arm position is

$$ {}^{base}T_{x\text{-ray}} = {}^{base}T_{tooth} \tag{8} $$

or

$$ {}^{base}T_{wrist}\ {}^{wrist}T_{x\text{-ray}} = {}^{base}T_{tooth} \tag{9} $$

$$ {}^{base}T_{wrist} = {}^{base}T_{tooth}\ {}^{wrist}T_{x\text{-ray}}^{\ -1} \tag{10} $$

where

$$ {}^{wrist}T_{x\text{-ray}}^{\ -1} = {}^{x\text{-ray}}T_{wrist} = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & -1 & 0 & 0 \\ 0 & 0 & -1 & L \\ 0 & 0 & 0 & 1 \end{bmatrix}. \tag{11} $$

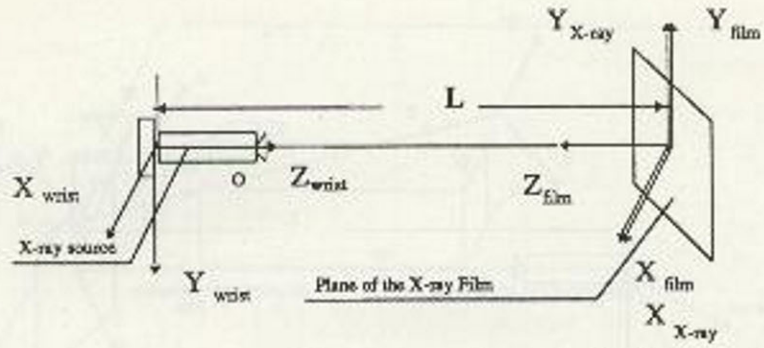

Fig. 4. Alignment of the X-ray source and targeted tooth plane.

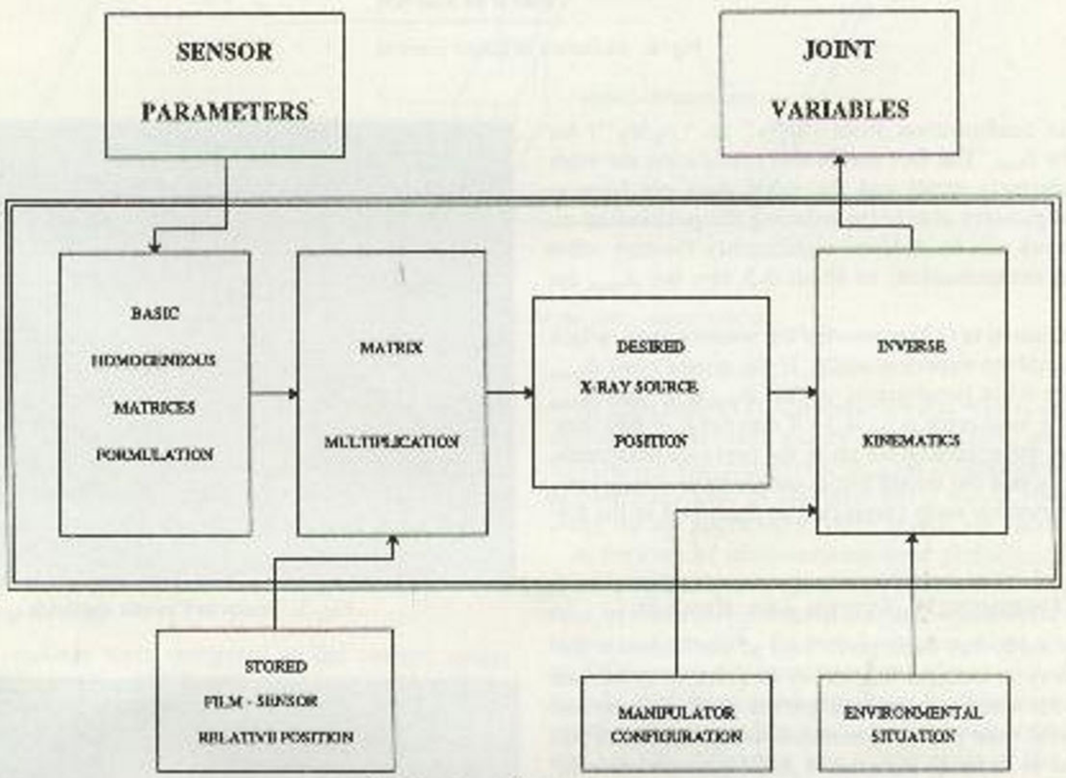

Fig. 5. Inverse kinematics process.

Equation (10) represents the solution which is fed into the robot controller. This in turn determines the robot joint positions based on inverse kinematics calculations [13]. The solution given by (10) is a static solution. In order to limit the adverse effects of the robot dynamics it is necessary to have a high bandwidth in the control loop (here the bandwidth of the sensor is the limiting factor), and relatively small and slow patient head motions. The overall process is shown in Fig. 5.

C. Solution Accuracy

The solution given by (10) is correct to the extent that there are no errors in the overall system. These errors, however, exist due to several factors, such as robot control inaccuracies, sensor-tooth relation, sensor errors, or errors in measurements that determine the fixed transformations ${}^{sensor}T_{tooth}$ and ${}^{base}T_{source}$. These factors compound the overall positioning error, so that $OX_{x\text{-ray}}Y_{x\text{-ray}}Z_{x\text{-ray}}$ will not coincide with $OX_{film}Y_{film}Z_{film}$. This is shown in Fig. 6. The upper bound for the correctable positioning error is the limit with which a projective invariant can be measured in the film as reported in the study by Dunn and van der Stelt [12] (XY translation errors of 0 to 16 mm and rotation errors of 0–32° about either the X, Y, or Z axis).

To a first approximation the total error Δ_{total} in the solution may be expressed as a function of robot translation error Δ_{tran}, robot rotation error Δ_{rot} and sensor translation error Δ_{stran} as

$$\Delta_{total} = \Delta_{tran} + L \tan(\Delta_{rot}) + \Delta_{stran}. \tag{12}$$

Here the effects of the sensor rotation errors (which are small) have been neglected, as well as the effect of the errors in measurements that determine ${}^{base}T_{source}$ and ${}^{sensor}T_{tooth}$. Special care needs to be taken to determine the fixed transformation ${}^{base}T_{source}$, by precise measurements of the robot and source system of coordinates. Δ_{tran} and Δ_{rot} are functions of the robot arm accuracy, which represents the difference between the commanded and actual positions of the robot. These errors vary from manipulator to manipulator and can be as large as 20 mm [8]

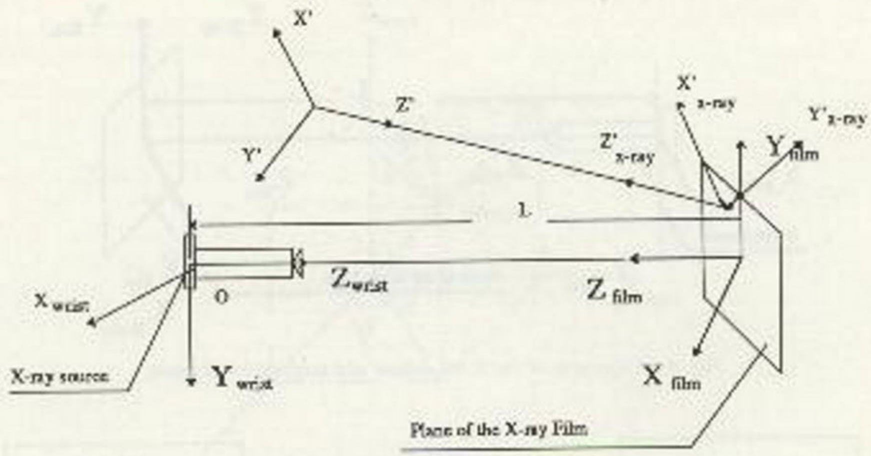

Fig. 6. Deviation of sensor position.

(for changes in configuration from "lefty" to "righty") for Δ_{trans} and 5° for Δ_{rot}. The fact that in this application the work envelope is relatively small and the robot does not have to change its configuration also helps reducing the positioning errors. These errors can be reduced significantly through robot calibration and compensation, to about 0.3 mm for Δ_{trans} for example [8].

The other unknown in (12) represents the sensor errors, which have to be determined experimentally. If the sensor error Δ_{sensor} is 3 mm and the robot is calibrated so that Δ_{rot} is 1° and Δ_{trans} is 1 mm then the total error Δ_{total} is 14.4 mm for $L = 600$ mm. This is less than the condition set up in the previous paragraph. In order to verify that the sensor errors satisfy these conditions, a series of experiments were conducted as described in the following section.

IV. Experimental System and Results

A feasibility study has been performed at the Robotics Research Laboratory at Rutgers University in order to determine the accuracy with which we can measure the tooth position and orientation in real time [7]. A sensorized mouth appliance has been developed in order to maintain a fixed geometrical relationship between sensor and tooth. The sensor-mouth appliance assembly is shown in Fig. 7. This appliance consists of a dental mold, a rigid plastic connecting piece attached to the mold, and a sensor attached to the connecting piece. The plastic connection is designed to allow the attachment of the radiographic film at one end, and that of the sensor at the other. The sensor seats outside the mouth and allows the patient to close his/her mouth. In this way the sensor may be used repeatedly for different patients and the only consumables are the plastic connection and the mouth appliance. The use of the plastic piece also allows the placement of the X-ray film at different locations in the mouth, which is a definite advantage over mechanical stents that allow the subtraction radiography of frontal teeth only. Of course, for different positions in the mouth there will be different connecting pieces. The connecting piece places the sensor away from the X-ray beam, so that the sensor wires and electronics are not shown in the image on the film.

The sensor utilizes low-frequency magnetic field technology to determine its position and orientation in relation to a source reference frame. It is capable to perform constant monitoring (max 60 times per second) of the position and orientation of an object in three-dimensional space [10]. The data from the magnetic sensor are sampled by an electronic unit that is also connected to the sensor source. This electronic unit is in turn interfaced with the host computer over an RS232 serial line as shown in Fig. 8. The computer is used for programming, data transfer, and analysis.

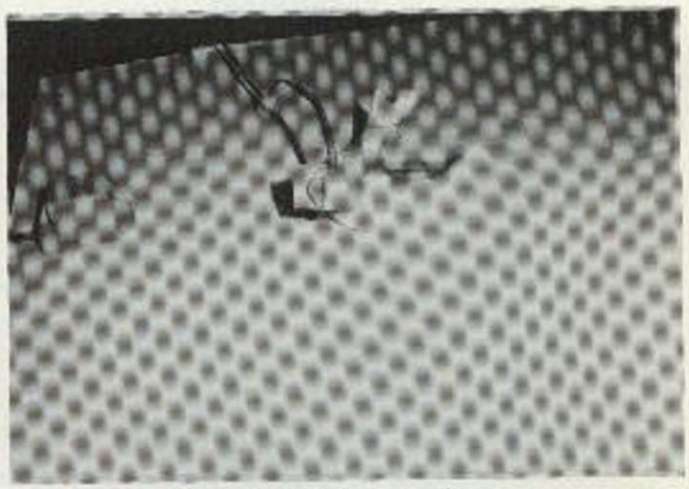

Fig. 7. Sensorized mouth appliance.

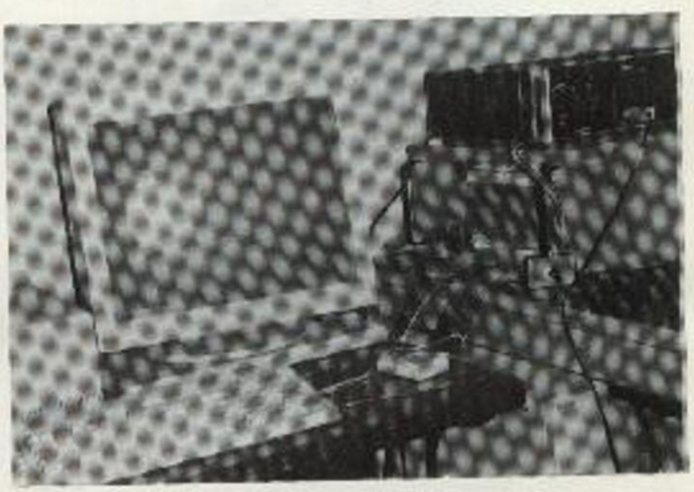

Fig. 8. Sensor interface and workstation.

The sensor and connecting piece were mounted on calibrated blocks and placed at known locations on a PVC (nonmetallic)

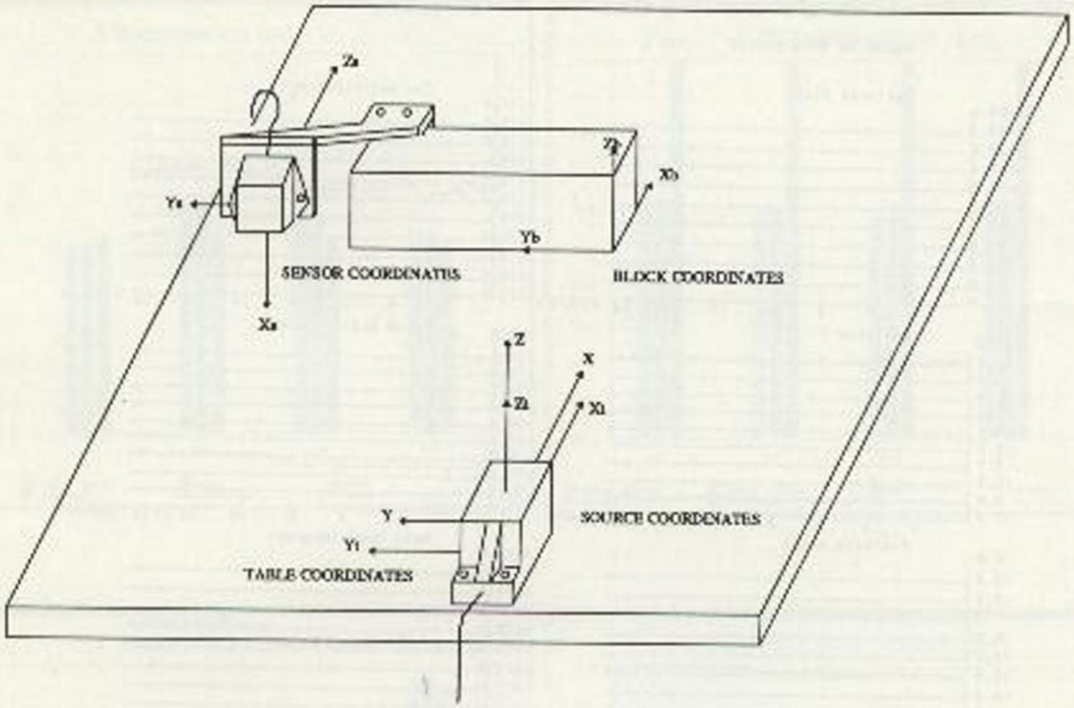

Fig. 9. Experimental table for accuracy measurements.

table. On the same table was rigidly mounted the sensor source, as shown in Fig. 9. The relationship between the block system of coordinates and the sensor system of coordinates was determined through measurements. This transformation was then used to determine the correct position of the sensor on the experimental table.

A. Experimental Results

The sensor readings were compared to the correct values measured off-line in order to determine sensor accuracy and repeatability. The data are shown in Fig. 10 for a stationary sensor (the solid line represents the correct data). In order to determine sensor noise as well as average position, a number of readings were made over a period of 11 s.

Subsequently the same series of tests were performed at different locations on the calibration table. These tests were made to determine if there is significant degradation of sensor measurements as the distance between sensor and source increases. The optimum sensor range was determined to be less than 300 mm from the source. This optimum distance corresponds to the distance that the sensor will have from the source in the dental applications, when the source is fixed on the back of the X-ray chair, or in some other convenient location. The results are presented in Fig. 11. When compared to the correct values, the sensor readings differ by at most 2.5 mm translations and 1° rotations.

The successful application of the above technique to dental subtraction radiography would require small translation errors. Our best results had sensor errors of less than 0.25 mm, while the largest sensor error was 2.5 mm. The larger errors are due to the use of a less accurate sensor which was available at the time of the experiments, as well as to a relatively imprecise calibration table. Thus, the "correct" values had a built-in reading error of 0.5–1.0 mm. According to its manufacturer, sensor measurements are not affected by X-rays.

As seen from (12), the experimental results on the sensor errors at optimal range satisfy the constraints set out in Section III. Under these conditions, Δ_{shift} is 2.5 mm or less for a sensor to source distance of 200 mm. Since this is within the optimal range for the application, these results are encouraging.

A third set of measurements were performed to determine if there is a significant degradation of sensor readings due to presence of metal (i.e., restorations and orthodontic appliances) in the patient's mouth. An initial set of readings were done for all six DOF with a dry mandible that had no metal. These readings showed that there is no difference in data when the skull was interposed between the source and the sensor block, or when it was not (the difference was on the order of the sensor noise). This is shown in Fig. 12. Subsequently, the same mandible was fitted with orthodontic wire and ten amalgam restorations and the same measurement procedure was applied. The data are presented in Fig. 13. These results are essentially the same as those shown in Fig. 12. The position readings differ from those shown in Fig. 12 by at most 0.1 mm. This indicates that there is no adverse influence on sensor accuracy due to the presence of metal in the patient's mouth.

V. Conclusions

A sensorized dental appliance that allows accurate measurement of targeted tooth position without supplemental mechanical alignment was described. The above described sensorized appliance avoids direct mechanical contact with the X-ray source and therefore can be used to image posterior as well as anterior teeth (mechanical techniques apply to anterior teeth only). Initial feasibility tests had encouraging results. These tests will be continued using a new generation sensor with better accuracy.

The long term goal of present research is to realize a complete system that integrates the robot manipulator with an X-

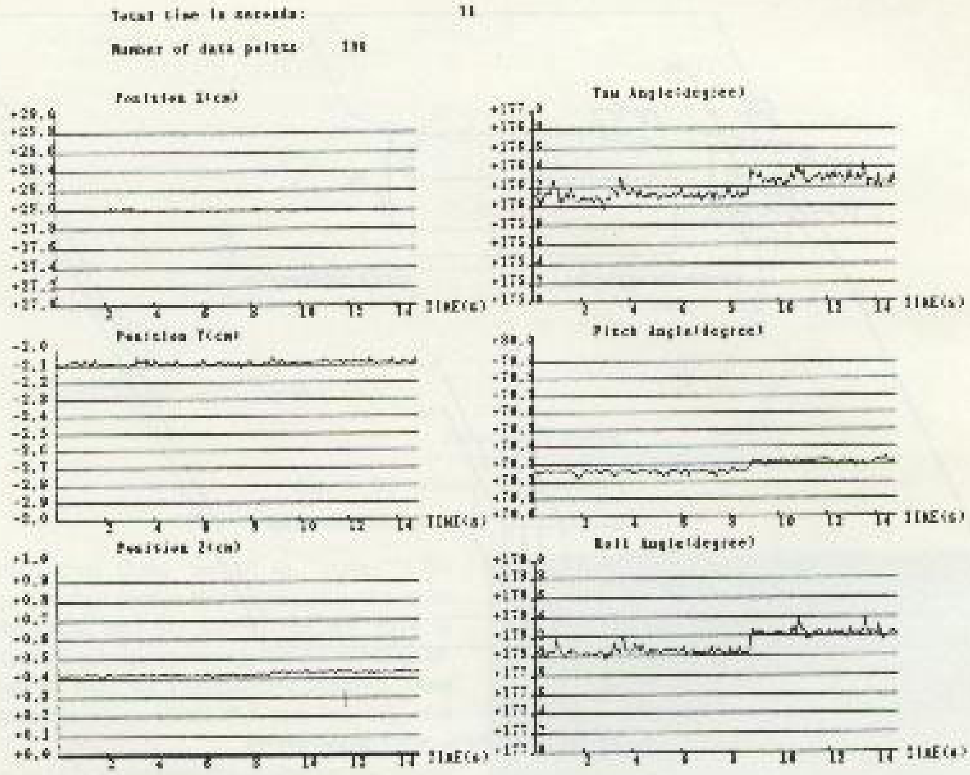

Fig. 10. Sensor readings (stationary).

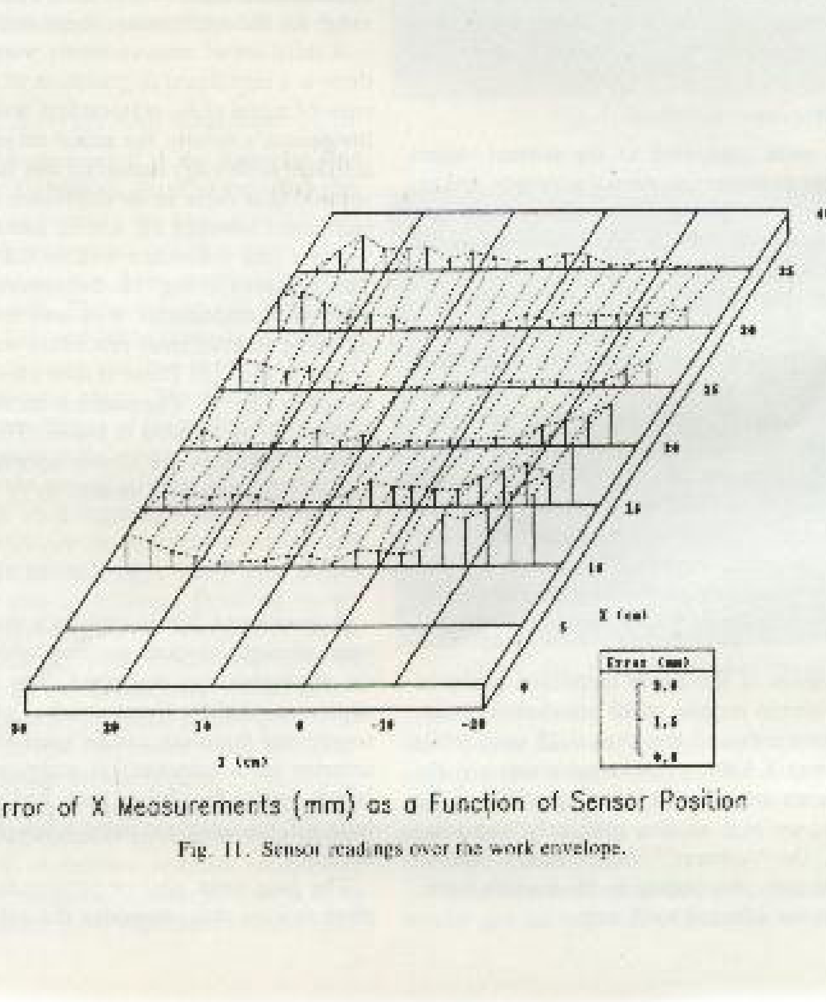

Fig. 11. Sensor readings over the work envelope.

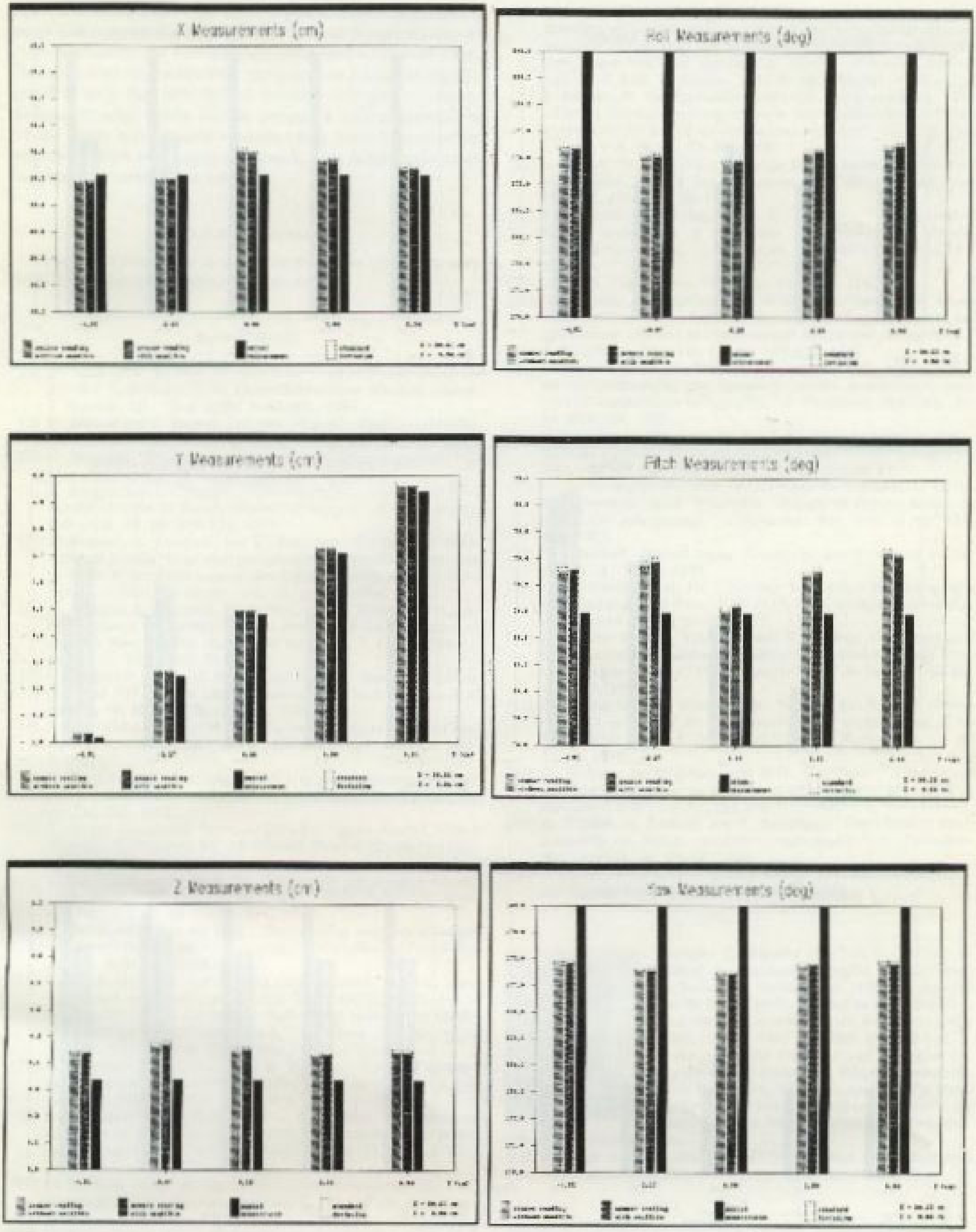

Fig. 12. Sensor readings from the dry mandible without metal.

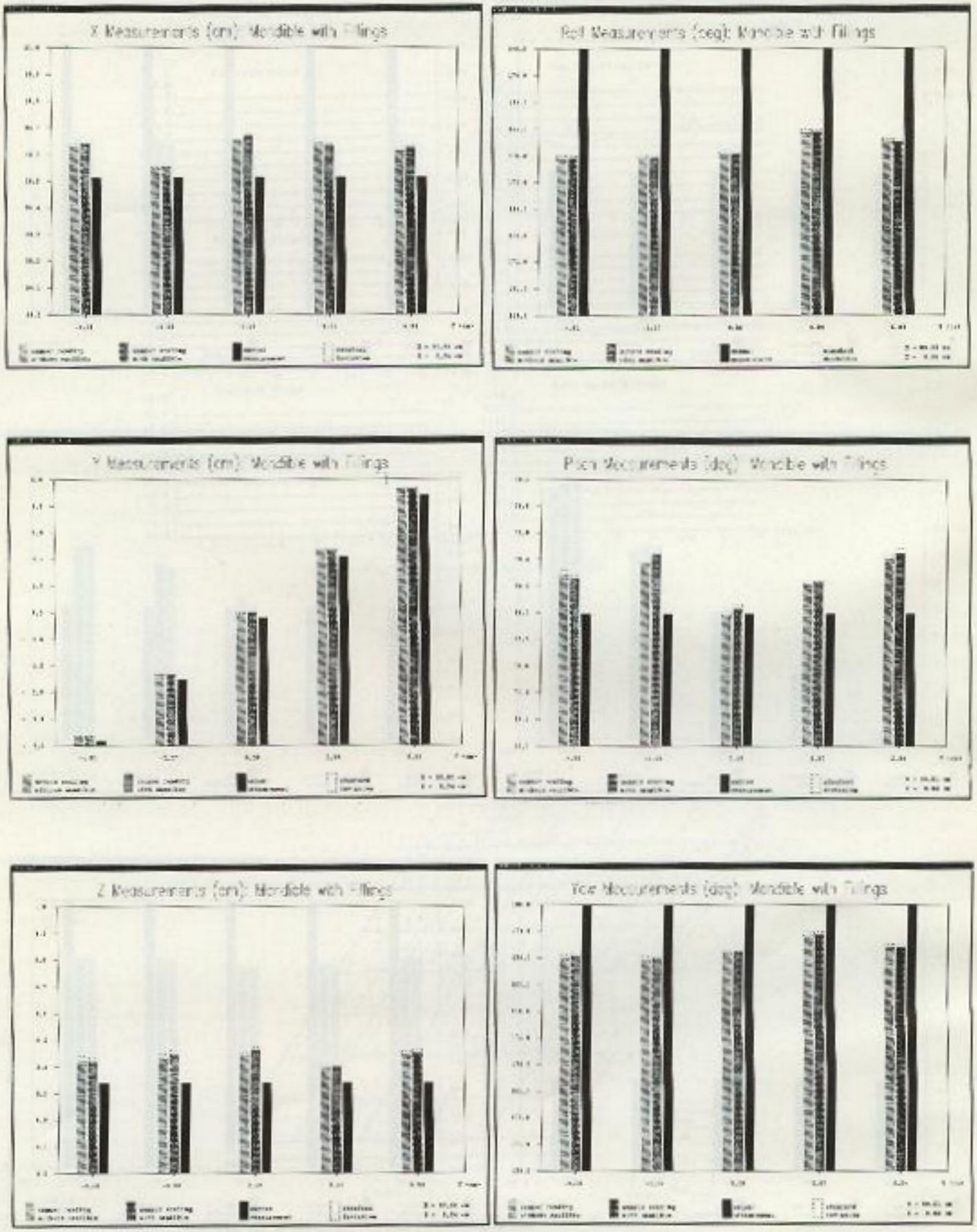

Fig. 13. Sensor readings from the dry mandible with metal.

ray source. The accuracy of this proposed system will be measured and compared with the model presented in this paper in order to identify and eliminate any additional sources of error, including bent film and errors introduced in fitting the mouthguard to teeth that have drifted between radiographic examinations. Finally, results will be compared with measurements of the same bone structure made using present mechanical alignment. Both techniques will benefit from image registration software commonly used today.

Acknowledgment

The authors would like to thank the reviewers who made very helpful suggestions in improving this paper.

References

[1] M. Altschuler, K. Bae, B. Altschuler et al., "Robot vision by encoded light beams," in *Three-Dimensional Machine Vision*, T. Kanade, Ed. New York: Academic, 1987.
[2] D. Ballard and C. Brown, *Computer Vision*. Englewood Cliffs, NJ: Prentice Hall, 1982.
[3] U. Bragger, "Digital imaging in periodontal radiography," *J. Clin. Periodont.*, vol. 15, pp. 551–557, 1988.
[4] U. Bragger and L. Pasquali, "Color conversion of alveolar bone density changes in digital subtraction images," *J. Clin. Periodont.*, vol. 16, pp. 209–214, 1989.
[5] U. Bragger, L. Pasquali, and K. Kornman "Remodeling of interdental alveolar bone after periodontal flap procedures assessed by means of computer assisted densitometric image analysis (CADIA)," *J. Clin. Periodont.*, vol. 15, pp. 558–564, 1988.
[6] U. Bragger, L. Pasquali, H. Weber, and K. Kornman, "Computer-assisted densitometric image analysis for the assessment of alveolar bone density changes in furcations," *J. Clin. Periodont.*, vol. 16, pp. 42–52, 1989.
[7] G. Burdea, S. Dunn, M. Mallik, and H. Jun, "Real time sensing of mouth 3-D position and orientation," in *Proc. SPIE Medical Imaging IV*, Newport Beach, CA, 1990.
[8] J. Chen and L. Chao, "Positioning error analysis for robot manipulation with all rotary joints," *IEEE J. Robot. Automat.*, vol. 3, 1987.
[9] J. Denavit and R. Hartenberg, "A kinematic notation for lower pair mechanisms based on matrices," *J. Appl. Mech.*, vol. 77, pp. 215–221, 1955.
[10] Polhemus Navigation Sciences Division, *Space Isotrak User's Manual*. Colchester, VT: McDonnell Douglas Electronics Co., 1987.
[11] S. Dunn, R. Keizer, and J. Yu, "Measuring the area and volume of the human body with structured light," *IEEE Trans. Syst. Man, Cybern.*, vol. 19, pp. 1350–1364, 1989.
[12] S. Dunn and P. van der Stelt, "Recognizing anatomic structure in arbitrary radiographic projections," *Dento Maxillo Facial Radiology*, to be published.
[13] K. Fu, R. Gonzalez, and C. S. G. Lee, *Robotics: Control, Sensing, Vision, and Intelligence*. New York: McGraw-Hill, 1987.
[14] H. Grondahl and K. Grondahl, "Subtraction radiography for diagnosis of periodontal bone lesions," *Oral Surg.*, vol. 55, no. 2, pp. 208–213, Nat. Inst. Dental Res., 1983.
[15] H. Grondahl, K. Grondahl, and R. Webber, "A digital subtraction technique for dental radiography," *Oral Surg.*, vol. 55, no. 1, pp. 96–102, Nat. Inst. Dental Res., 1983.
[16] K. Grondahl, H. Grondahl, and R. L. Webber, "Influence of variations in projection geometry on the detectability of periodontal bone lesions," *J. Clin. Periodont.*, vol. 11, pp. 411–420, 1984.
[17] G. Herman, *Image Reconstruction From Projections. The Fundamentals of Computerized Tomography*. New York: Academic, 1980.
[18] B. Klaus and P. Horn, *Robot Vision*. Cambridge, MA: MIT Press, 1986.
[19] P. Janssen, W. van Palenstein Helderman, and J. van Aken, "The detection of in vitro produced periodontal bone lesions by conventional radiography and photographic subtraction radiography using observers and quantitative digital subtraction radiography," *J. Clin. Periodont.*, vol. 16, pp. 335–341, 1989.
[20] P. Janssen, W. van Palenstein Helderman, and J. van Aken, "The effect of in-vivo-occurring errors in the reproducibility of radiographs on the use of the subtraction technique," *J. Clin. Periodont.*, vol. 16, pp. 53–58, 1989.
[21] R. Jarvis, "A perspective on range finding techniques for computer vision," *IEEE Trans. Pattern Anal. Mach. Intell.*, vol. PAMI-5, pp. 122–139, 1983.
[22] M. Jeffcoat, R. Jeffcoat, and R. C. Williams, "A new method for the comparison of bone loss measurements on nonstandardized radiographs," *J. Periodont. Res.*, vol. 19, pp. 434–440, 1984.
[23] D. Marr, *Vision*. San Francisco: Freeman, 1982.
[24] K. McHenry, E. Hausmann, U. Wikesjo, R. Dunford, E. Lyon-Bottenfield, and L. Christersson, "Methodological aspects and quantitative adjuncts to computerized subtraction radiography," *J. Periodont. Res.*, vol. 22, pp. 125–132, 1987.
[25] L. Ortman, R. Dunford, K. McHenry, and E. Hausmann, "Subtraction radiography and computer assisted densitometric analyses of standardized radiographs," *J. Periodont. Res.*, vol. 20, pp. 644–651, 1985.
[26] J. Posdamer and M. Altschuler, "Surface measurement by space encoded projected beam systems," *Comput. Graph. Image Process.*, vol. 18, pp. 1–17, 1982.
[27] M. Rethman, U. Ruttiman, R. O'Neal, R. Webber, A. Davis, G. Greenstein, and S. Woodyard, "Diagnosis of bone lesions by subtraction radiography," *J. Periodont. Res.*, vol. 56, pp. 324–329, 1985.
[28] R. Schalkoff, *Digital Image Processing and Computer Vision*. New York: Wiley, 1989.
[29] G. Stockman and G. Hu, "Sensing 3-D surface patches using a projected grid," in *Proc. 1986 IEEE Conf. Comput. Vision Pattern Recogn.*, 1986, pp. 602–607.
[30] P. van der Stelt, U. Ruttiman, and R. Webber, "Determination of projections for subtraction radiography based on image similarity measurements," *Dento Maxillo Facial Radio.*, vol. 18, pp. 113–117, 1989.
[31] P. van der Stelt, U. Ruttiman, R. Webber, and R. A. J. Groenhuis, "A procedure for reconstruction and enhancement of tomosynthetic images," *Dento Maxillo Facial Radio.*, vol. 15, pp. 11–18, 1986.
[32] R. Webber, U. Ruttiman, and H-G. Grondahl, "X-ray image subtraction as a basis for assessment of periodontal changes," *J. Periodont. Res.*, vol. 17, pp. 509–511, 1982.
[33] R. Webber, A. Tzukert, and V. Ruttiman, "The effects of beam hardening on digital subtraction radiography," *J. Periodont. Res.*, vol. 24, pp. 53–58, 1989.
[34] I. Weiss, "Projective invariants of shapes," in *Proc. Image Understanding Workshop*, 1988, pp. 1125–1134.

Grigore C. Burdea (S'87–M'87–SM'91) received the engineering degree (valedictorian) in Bucharest, Romania, in 1980. He received the M.S. and Ph.D. degrees in applied science and robotics from New York University, New York City, in 1985 and 1987, respectively.

He joined the Electrical and Computer Engineering Department at Rutgers University, New Brunswick, NJ, as an Assistant Professor in January, 1988. His research interests are force feedback control, human-machine interfaces and medical robotics.

Dr. Burdea currently chairs the IEEE Robotics and Automation Chapter-Princeton Section.

Stanley M. Dunn (S'78–S'84–M'84–M'85), for a photograph and biography, see this issue, p. 313.

Charles H. Immendorf is working towards the B.S. degree in electrical and computer engineering at Rutgers University, New Brunswick, NJ. His interests are computer graphics and software engineering.

Madhumita Mallik received the B.S. degree from Regional Engineering Collage, Durgapur, India, in 1989.

She is currently working towards the M.S. degree in electrical and computer engineering at Rutgers University, New Brunswick, NJ, in May 1991. Her interest are robotics and VLSI design.