

The Development and Clinical Application of a Patient Position Monitoring System*

L.H. Gerig¹, S.F. El-Hakim², J. Szanto¹, D. Salhani¹ and A. Girard¹

¹Ottawa Regional Cancer Centre, General Division, 501 Smyth Rd., Ottawa, Ontario, Canada K1H 8L6

²National Research Council, IIT, Ottawa, Ontario, Canada K1A 0R6

ABSTRACT

We have developed and clinically tested a computer vision system capable of real time monitoring of the position of an oncology (cancer) patient undergoing radiation therapy. The system is able to report variations in patient setup from day to day, as well as patient motion during an individual treatment. The system consists of two CCD cameras mounted in the treatment room and focused on the treatment unit isocentre. The cameras are interfaced to a PC via a two channel video board. Special targets, placed on the patient surface are automatically recognized and extracted by our 3-D vision software. The three coordinates of each target are determined using a triangulation algorithm. System accuracy, stability and reproducibility were tested in the laboratory as well as in the radiation therapy room. Beside accuracy, the system must ensure the highest reliability and safety in the actual application environment. In this paper we also report on the results of clinical testing performed on a total of 23 patients having various treatment sites and techniques. The system in its present configuration is capable of measuring multiple targets placed on the patient surface during radiation therapy. In the clinical environment the system has an accuracy and repeatability of better than 0.5 mm in Cartesian space over extended periods (> 1 month). The system can measure and report patient position in less than 5 seconds. Clinically we have found that the system can easily and accurately detect patient motion during treatment as well as variations in patient setup from day to day. A brief description of the system and detailed analysis of its performance in the laboratory and in the clinic are presented.

Key Words: Position monitoring, Radiology, Oncology, Stereo vision, Target tracking, Safety critical software.

1. INTRODUCTION

The accurate placement and verification of treatment fields remains one of the principal problems in the delivery of radiation therapy. Measurements of patient setup and field positioning errors have been reported by many authors, with mean deviations of the order of 5 to 8 mm and a significant percentage of errors in excess of 15 mm.^{1,2,3,4,5,6} The decrease in Tumour Control Probability (TCP) resulting from such errors has been discussed by several authors and can be as large as 20%.^{7,8,9,10}

Recently, many groups have attempted to address this issue through the application of port films or real time portal imaging technology.^{11,12,13-15,16,17,18} Presently the routine clinical implementation of these approaches remains labor intensive, requires human judgment, and the delivery of radiation before patient/field position can be determined. Further, most reporting and analysis is done off-line after patient treatment has been completed.

To date, there is no commercial system which can rapidly, reliably, remotely and accurately measure the orientation/position of a patient in Cartesian space. In this work we report the development and clinical application of an optically based system capable of detecting patient setup position and motion during treatment.

2. SYSTEM DESCRIPTION

The Patient Position Monitoring (PPM) system described herein, is presently being developed at the Ottawa Regional Cancer Centre with financial support from Siemens Medical Laboratories - Oncology Care Systems of Concorde, California. It is a computerized video based system capable of measuring the Cartesian coordinates of small optical targets placed on the surface of a patient. When the patient moves between measurements, the system determines the change in target position and reports a change in patient position. If the targets are placed in the same position on the patient from day to day, the reproducibility of the patient setup can be determined. There are several significant advantages to such a system. The position of the targets or markers (and by inference the patient) can be accurately determined before radiation is delivered. The

* in Videometrics III, Proc. SPIE 2350, Boston, MA, Nov. 2-4, 1994

system can determine the relative motion (setup position) of the patient from day to day, and unlike real-time portal imaging, it is sensitive (better than 0.1 mm) to motion in any of the 3 principal axes and to patient rotation in any plane.

2.1. System Requirement Specifications

The design criteria used for the development of the system included the following specifications:

- Measure and report the position of up to 10 targets every 5 seconds.
- Recognize and identify all targets within a volume of space defined by a sphere of 30 cm diameter, whose centre is placed in the vicinity of the therapy unit isocentre.
- Resolve the absolute target position to better than 1 mm in Cartesian space and determine patient rotation about any of the three principal axes.
- Target design such that a technologist could reasonably be expected to replace the targets on the patient with a reproducibility of better than 2 mm.
- The ability to warn the operator/technologist if the patient (targets) are not in the correct position before treatment.

2.2. System Design

The vision system chosen for this application; the vision-based coordinate measurement (VCM) system, is capable of achieving an accuracy of 0.01 mm over the volume required for this application.²⁰ However, to achieve this accuracy, stringent requirements must be met. The use of precise target center algorithms, taking several images and averaging, and correcting certain errors such as CCD line-jitters, to name just a few, have to be implemented. These requirements can be computationally costly and may require special training and restrictions on the use of the system. Since the specified accuracy for this application is 1 mm, these stringent requirements may be relaxed and the emphasis is now shifted from achieving the best possible accuracy to ensuring the highest possible cost-efficiency, reliability, speed, safety, and ease of use in the actual application environment.

2.2.1. Hardware Design

The hardware consists of a 486 personal computer, a high resolution graphics display, an image acquisition and processing board, two CCD cameras with 25 mm lens and two diode lasers (810 nm wavelength) with dispersion lens, one laser mounted immediately on top of each camera (figure 1). The camera lenses are fitted with 810 +/-5 nm band pass filters. The cameras and lasers are oriented with respect to the patient as shown in figure 2, such that the cameras are 1.6 m above isocentre, 1.0 m either side of the gantry axis of rotation and 1.6 m back from the isocentre. The targets, which are placed on the patient, are constructed of a retroreflective material, such that the 810 nm infrared light originating from atop of each camera is reflected back to the camera through the band pass filter providing a high degree of inherent target contrast.

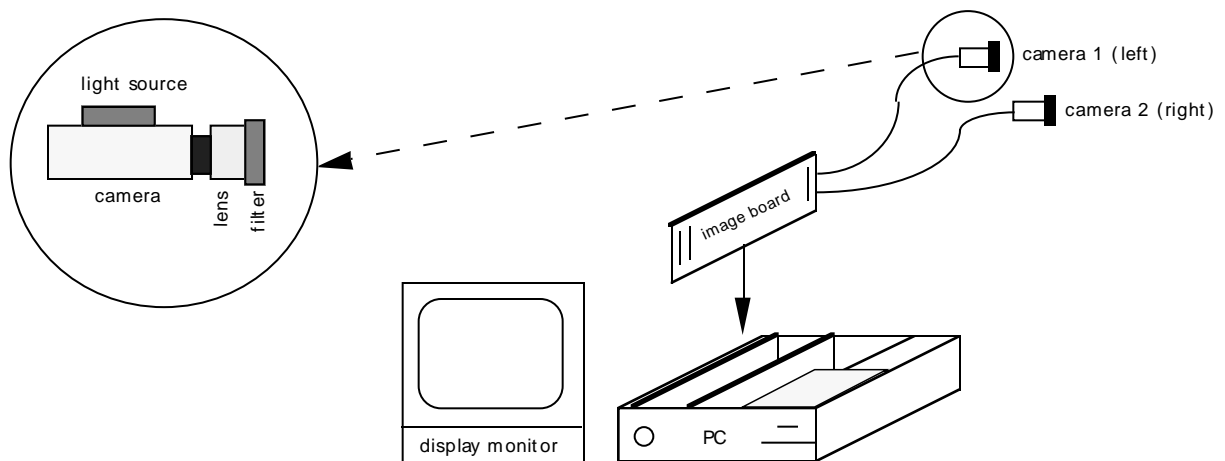


Figure 1: Main hardware components of PPM

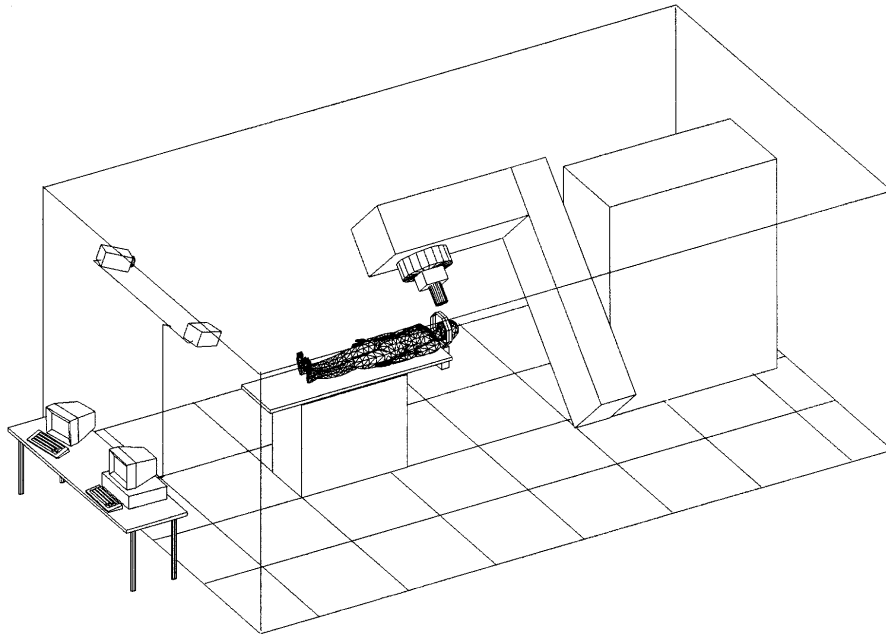


Figure 2: Patient / camera position

2.2.2. Software design

The PPM software consists of two basic components, the 3-D vision software (VCM) to recognize and measure the position of the targets, and the user interface and analysis software (figure 3).

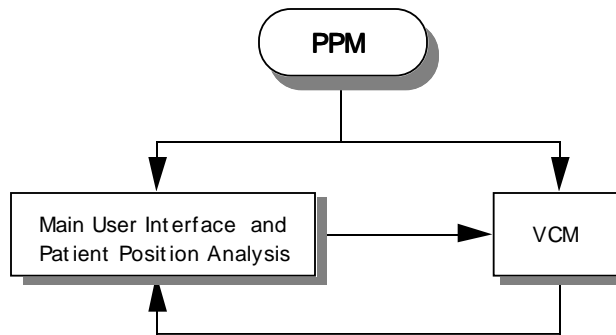


Figure 3: Software components of PPM

2.2.2.1. Vision Software

The VCM system has been under development at the National Research Council of Canada for several years.^{19,20} The system is a software package which can be integrated with commercially available solid-state cameras, image grab and processing boards, and computer hardware. It combines the principles of stereo vision, photogrammetry and knowledge-based techniques to provide precise coordinate and dimension measurements of objects for a diverse range of applications. The VCM system has been used for other medical applications including the measurement of orthopedic seats and anatomical stumps.²¹ As applied in the PPM, the VCM system is used to identify and measure the position of targets placed on the patient surface. Figure 4 summarizes the inputs and outputs of the system. The VCM software consists of a set of library routines which are utilized to design application programs and combine with application-specific user interface. These library functions are dynamic link libraries (DLL) written in C for Microsoft Windows.

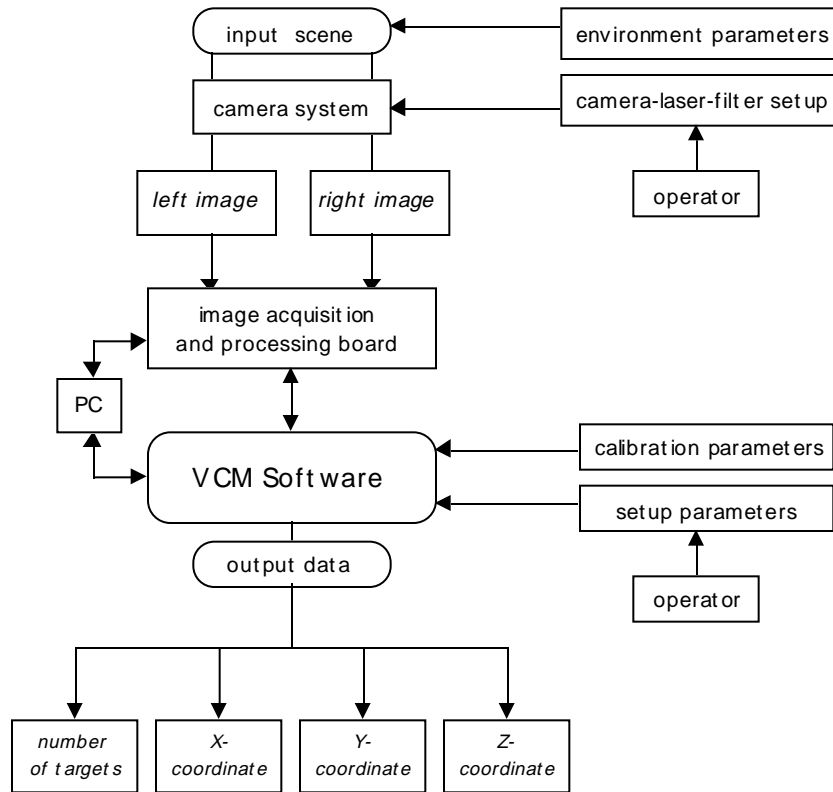


Figure 4: Inputs and outputs for VCM software

Rigorous camera calibration is essential for vision applications, particularly when accurate measurements are required and to ensure successful stereo matching. The objective of the calibration is to determine the position and orientation of the camera with respect to a selected coordinate system, the focal length and coordinates of the principal point on the image, and several additional parameters to compensate for distortions produced by the lens, pixel size and alignment, and other sources. The model is well documented in photogrammetry literature^{22,23} and will not be described here. The calibration requires points of precisely known positions in the object space coordinate system. A set of well-defined targets mounted at various heights on one side of a thermally stable block, the calibration plate (jig), is employed (figure 5(a)). The plate is placed facing the cameras, at about 45° angle with the treatment table (figure 5(c)). A set of 4 targets arranged in a square (figure 5(b)) are placed directly on the table and used to transfer the calibration parameters from the calibration-plate coordinate system to the treatment-table coordinate system.

The target extraction and measurement process has been described in detail elsewhere.¹⁹ It consists of several hierarchical steps (figure 6). The segmentation processing reduces the gray scale image into a binary image with an automatic thresholding routine. Targets are isolated by performing connectivity analysis to separate images into blobs. Blob parameters such as area, perimeter, and radius are used to identify which image components qualify as the targets that should be measured. Sub-pixel target locations are achieved by computing the blob centroid. Target locations from different images are then matched using selected constraints to aid in the matching process. Once targets have been successfully matched, the calculation of their 3-D real-world locations can be made by using the calibration parameters in a triangulation process.

The iterative, hierarchical matching process is controlled by sets of constraints determined by geometric relationships and a priori knowledge, as follows:

- (1) The epipolar line: To match a point in one image with its corresponding point in the other, the image coordinates of the point in the first image and the calibration parameters of the two images are used to determine the relationship between x and y coordinates of the corresponding point in the second image. This is a straight-line equation, and the image coordinates of all recognized points in the second image are tested to determine which one falls on this line.

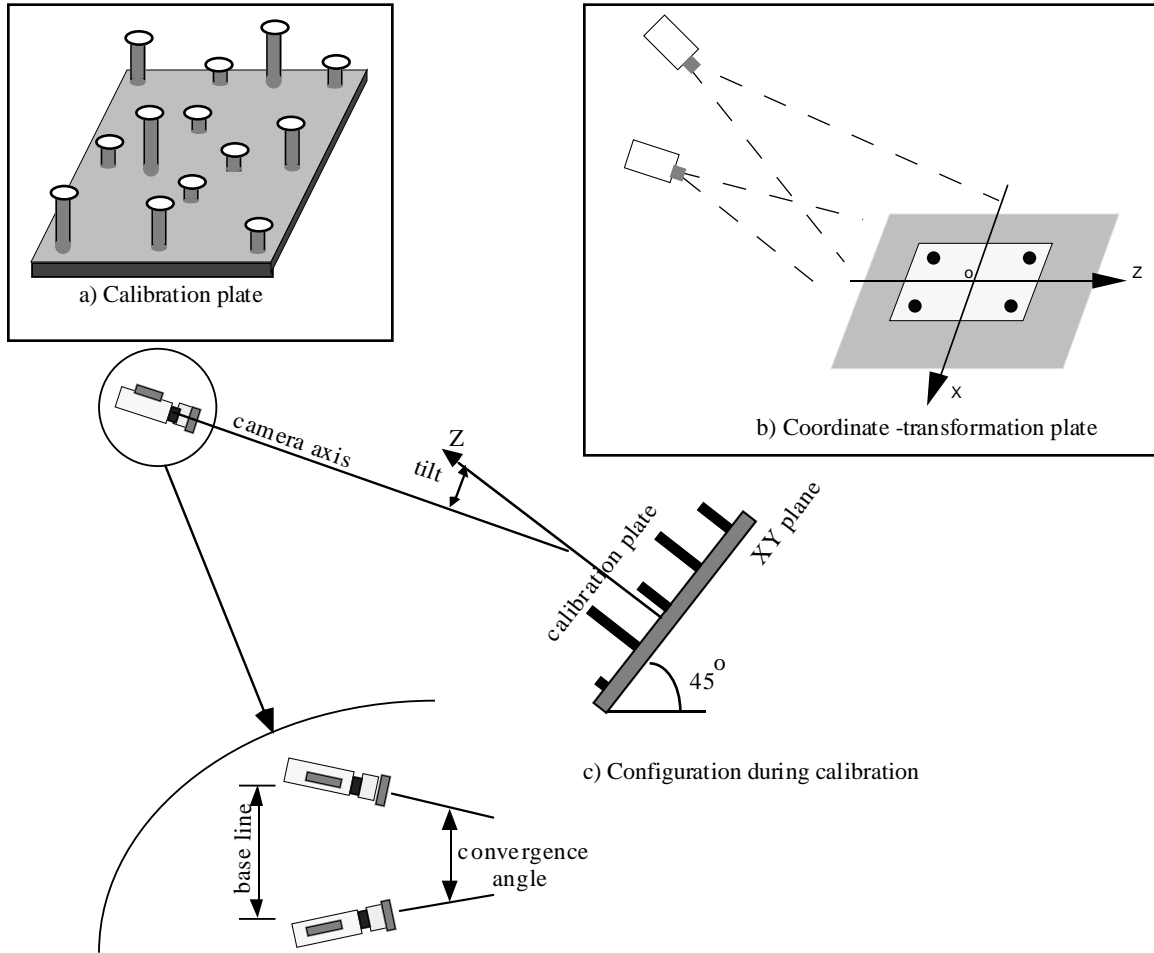


Figure 5: PPM calibration

(2) The disparity constraint: The expected range of disparity (the difference between the x-coordinates from the 2 images) is computed from two sources; the depth constraint using the known a priori expected range of depth as determined from the user defined initial setting procedure, and the already successfully matched points in earlier iterations.

(3) If more than one point satisfies the above constraints, an ordering constraint is applied.

These constraints are input into the VCM software during the setup step (part of figure 4) as part of the installation process and should not require any change afterwards.

2.2.2.2. User Interface and Analysis Software

The user interface has been developed in Microsoft Visual Basic to run under Microsoft Windows 3.1. It is a menu driven system which provides drop down windows designed to afford the user with the maximum flexibility and ease of use. The system provides a data base for patient demographics and storage of patient position data.

Objects measured by the VCM software have their position reported in the object space defined by the calibration points. This is converted to the conventional therapy unit coordinate system by a simple transformation matrix, such that the isocentre is defined as "0,0,0", the X axis lies in the horizontal plane perpendicular to the gantry axis of rotation, the Z axis is coincident with the gantry axis of rotation and the Y axis is mutually perpendicular to the other two axes and defines the height with respect to isocentre. The transformation matrix is determined by the VCM measurement of a transformation jig aligned exactly with the principal axes of the therapy unit and centered at the isocentre (figure 5(b)).

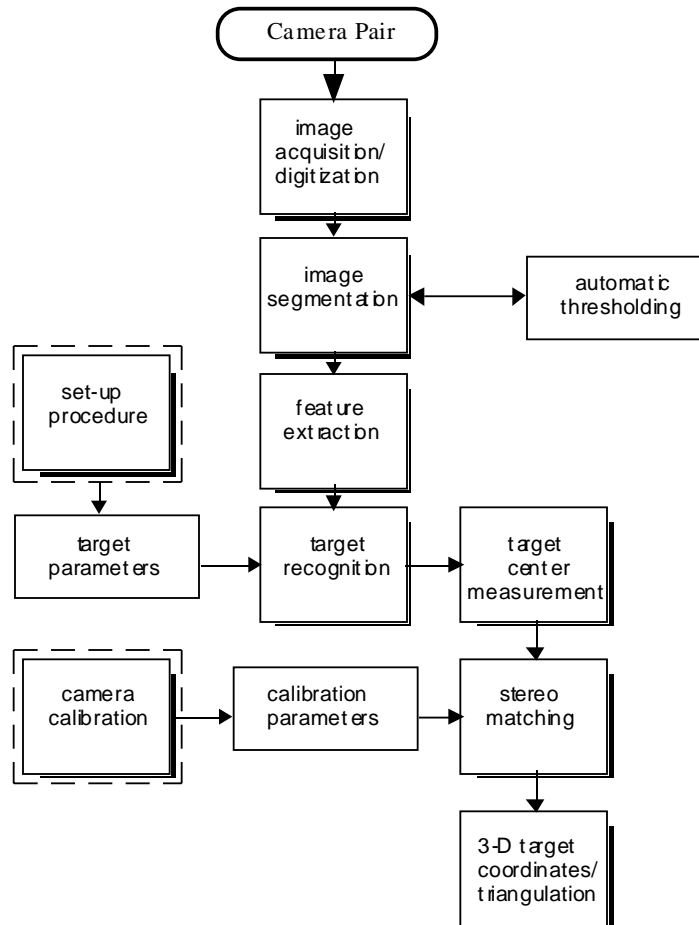


Figure 6: Vision processing steps

In routine use, the technologist places three to five circular (15.0 mm diameter) self-adhesive, retroreflective paper targets on the patient's surface which do not move during the course of each fraction (treatment cycle). Targets may be left on the patient until their retroreflective surface becomes degraded (usually four to five days) or can be removed after each treatment and replaced the following day. When targets are placed on a patient immobilization device such as a shell, there is no requirement to replace them during a normal treatment course. The centre of the targets have a 2.0 mm diameter hole, which provides the mechanism by which the technologist can easily and accurately relocate the targets at the same place (mark) on the patients surface from day to day. Placement of the targets requires approximately 15 to 30 seconds per patient depending upon the number of targets used and the location on the patient.

The PPM system measures and uses three types of patient position data: reference position, setup position and treatment position. The reference data are the coordinates of each target in Cartesian space, measured at the time of the patient's initial treatment setup when the check films are taken or at simulation. There is one set of reference data for each treatment isocentre. Setup and treatment data are reported relative to the reference position. Measurement of the treatment setup position provides the opportunity to determine and adjust the patient position prior to the delivery of radiation. Treatment position (i.e. deviation from the reference position) is measured and reported every five seconds while the therapy beam is on. PPM systems can be installed in the simulator suite and therapy units, each are calibrated to assure identical coordinate systems. This then provides the mechanism whereby patients can be transferred between units while still using the same target position information.

The patient's position with respect to the reference condition is calculated such that the deviation in the X direction, X_{dev} , is given as:

$$X_{dev} = \frac{\sum_{i=1}^N X_{oi} - X_i}{N} \quad (1)$$

where X_{oi} is the absolute reference X position of the i th target and X_i is the absolute X position of the i th target at the time of sampling. Deviations in the Y direction and the Z direction are calculated in a similar manner. The total deviation R_{dev} is calculated directly as:

$$R_{dev} = \sqrt{X_{dev}^2 + Y_{dev}^2 + Z_{dev}^2} \quad (2)$$

In order to implement equation 1, it is necessary to be able to uniquely identify each target. The process of target identification ensures that the targets placed on the patient for a specific treatment can be identified with a 1:1 matching to the targets measured when the reference position was determined. Our target identification algorithm performs two functions. It discards extraneous targets reported by the VCM software and matches the targets measured at treatment time to those measured at the time the reference setup was established.

The technique of uniquely matching identified target at treatment or setup with targets measured at reference is as follows. Let $r_{i,j}$ be the vector joining the i th and j th target identified when the reference position was measured. Similarly, let $v_{k,l}$ denote the vector joining the k th and l th targets measured at treatment time. Target identification depends solely on the correspondence of those relative vectors with each other. That is to say, if $r_{i,j}$ and $v_{k,l}$ are equal then it follows that target k must correspond to target i and similarly l to j . The equivalence of the relative vectors are assumed if:

$$\begin{aligned} & \|r_{i,j} - v_{k,l}\| < \varepsilon_1 \\ \text{and} & \\ & \|\psi\| < \varepsilon_2 \end{aligned} \quad (3)$$

where ε_1 and ε_2 are prescribed target placement tolerance, and ψ represents the angle between the vectors $r_{i,j}$ and $v_{k,l}$. The angle ψ is determined by:

$$\begin{aligned} \psi &= \pi - \cos^{-1}\left(\frac{r_{i,j} \cdot v_{k,l}}{\|r_{i,j}\| \|v_{k,l}\|}\right) & r_{i,j} \cdot v_{k,l} < 0 \\ \text{or} & \\ \psi &= \cos^{-1}\left(\frac{r_{i,j} \cdot v_{k,l}}{\|r_{i,j}\| \|v_{k,l}\|}\right) & r_{i,j} \cdot v_{k,l} > 0 \end{aligned} \quad (4)$$

It should be noted that the sense of the relative vectors is important in determining target coincidence. The target identification algorithm is fairly robust, however, it will fail under certain circumstances. For situations where the targets have been placed in an axially symmetric pattern the algorithm could report ambiguous results. Placement of the targets in an asymmetric pattern will preclude this problem.

Patient rotations are specified as tilt, yaw and roll, which are defined as rotations about the axes parallel to the principal X, Y, Z axes respectively. These are the mean angular deviations of the targets, relative to their reference position, which are determined from the difference in the angular position of each measured target from the position of the corresponding reference targets. The angular position of each reference and measured target is calculated about the centre of mass of the reference targets and measured targets respectively, where only those targets uniquely identified by the target identification algorithm described above are included in the calculation. Thus, patient *Yaw* is specified as:

$$Yaw = \sum_{i=1}^N \frac{(\theta_{oi} - \theta_i)}{N} \quad (5)$$

where

$$\theta_{oi} = \tan^{-1} \frac{(Z_{oi} - Z_{ocm})}{(X_{oi} - X_{ocm})} \quad (6)$$

and Z_{ocm} and X_{ocm} are the Z, X coordinates of the centre of mass of the reference targets. θ_i is given similarly using its own centre of mass as reference.

In general the centre of mass of the targets does not lie on the axis of rotation of the therapy couch. As a result, there are actually two sets of deviations which can be reported. The first are the deviations in X, Y, Z and tilt roll and yaw as defined above. However, the direct application of these corrections to the patient position, assuming that yaw is corrected by a couch rotation, would not move the patient back to the desired treatment position. In order to use the couch rotation to correct setup or treatment position, a transform matrix must be calculated from Yaw, X_{dev}, Z_{dev} to correctly move the patient back to the reference position using only the linear and rotational couch motions. Since there are no couch motions to correct for tilt and roll, it is assumed that these will be corrected by rotation of the patient about axes passing through the centre of mass of the targets.

Patient treatment and setup position are displayed to the therapist using the screen shown of Figure 7. Real time deviations in displacement (X, Y, Z , and R) and the rotations about the three principal axes are displayed in the upper left and upper right quadrants of the user display respectively. The deviations are presented numerically and as coloured bars whose lengths are proportional to the magnitude of the deviation. The bars are colour coded such that displacements (or rotations) which exceed the tolerances requested by the radiation oncologist are displayed in red, deviations which are greater than some percentage (e.g. 80%) of the allowed tolerance are displayed in yellow as a warning and deviations less than that are displayed as green. Both the tolerances (red) and warning level percentage are user definable and selectable at run time from a list.



Figure 7: Patient position display used by the therapist during treatment

Historical data for the patient in question is shown in the bottom left and right of the screen display. The bottom left quadrant is a time history of the magnitude of the displacement vector R for the current field and fraction. The bottom right graph is a history of R_{av} , the magnitude of the vector R averaged over each treatment field/fraction.

2.3. System Safety and Fault-Tree Analysis

Since this software will be used with a medical device, operational safety is an important factor. Means of minimizing hazards and means for minimizing damage in case of failure (failure tolerance) must be provided. System safety analysis is the starting point for system concept and continues throughout the design, implementation, verification and validation (V&V) and, maintenance cycles of the system. Preliminary hazard analysis is best carried out using the overall system fault tree (FT). Fault tree analysis (FTA) is one of the methods used for performing most of the V&V tasks required for safety critical software.²⁴ The fault tree is a graphic model of various parallel and sequential combinations of faults that will result in the occurrence of a predefined undesired event. The goal here is to design the system so that those faults are prevented and to show that the system is safe both if it operates as intended and in the presence of faults.

The first step is to identify the undesired event, or the hazard, and then work backwards to find all combinations of faults that produce this hazard. Safeguards against those faults and actions in case they occur will be embedded in the design, both in software and hardware. The potential hazardous event resulting from relying on the PPM system in positioning and monitoring the patient during oncology treatment is that the treatment is applied to the wrong body area (good tissue instead of the cancerous tissue). Two possibilities may cause such event (Figure 8):

(1) The system reported erroneously that the patient movement exceeded the acceptable limits. As a result, the patient was moved back by the erroneously-reported amount and the radiation was applied to the wrong body-area.

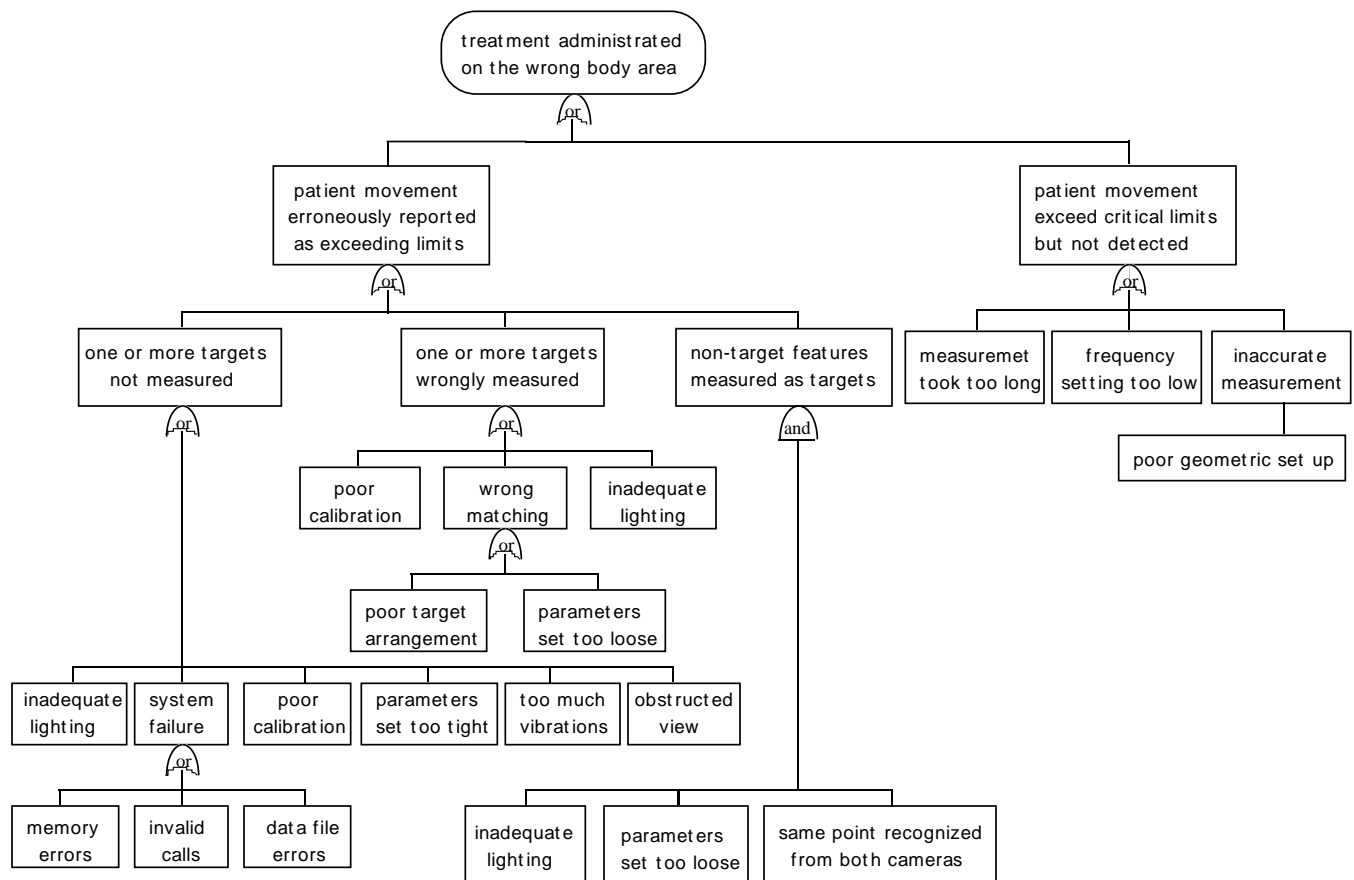


Figure 8: Fault Tree Analysis (FTA) for the PPM system

(2) The patient actually moved more than the acceptable limit, however the system did not sense such movement and did not alert the radiation therapist. This will result in continuing the treatment to the wrong body-area.

The possible causes for these two faults are shown in figure 8.

3. IMPLEMENTATION AND TEST RESULTS

The performance of the PPM system has been assessed under controlled laboratory conditions and in the clinical environment over an extended period of time.

3.1. Laboratory Results

The laboratory tests were performed to determine the optimal performance conditions of the VCM software and to assess its performance and robustness under optimal and suboptimal conditions. The coordinate system for laboratory measurements is defined such that X is the vector between the two cameras, Y is mutually perpendicular to X and the plane containing the central axis of vision and the two cameras while Z is mutually perpendicular to X and Y.

It is important to evaluate the system accuracy under optimal conditions in order to obtain the accuracy of the measurement tool itself rather than the influence of external factors. For this a three-axes (XYZ) positioning stage was employed as an independent superior measurement system for qualitative assessment of the vision-system measurements. In this work only results relative to target measurement accuracy are summarized. Complete test results can be found in earlier publications.²⁰

In order to ensure random sampling when repeated measurements are required to compute the mean and standard deviation, realistic variations in conditions must be used. It has been shown²⁵ that four or more independent measurements must be made in order to have a representative sample. The vision system bias is computed as the difference between the mean of the measured value and the value obtained by the independent superior system. Specifically, the laboratory tests have the following objectives:

- (1) Establish the best geometrical configuration for the given application setup.
- (2) Determine the statistical stability (precision) of the measurements under a variety of conditions.
- (3) Determine the measurement bias relative to a superior measurement system.

The laboratory tests have assessed the metric performance of the VCM system and determined the geometric conditions required for optimal performance. These results can be summarized as:

- To obtain the optimal performance, the ratio between the camera's baseline distance and the object average standoff distance should be at least 0.7, translating to a convergence angle between the two cameras of at least 38.6° .

- Repeatability results under the tested parameter variation gave a standard deviation of 0.014 mm and a maximum deviation of 0.038 mm, which significantly exceeds the requirements of all clinical applications. In general the measurement tool should have a precision and accuracy at least an order of magnitude better than the accuracy with which the unknown parameter is to be measured.

- The measurement bias was found to be significantly larger for targets lying near the image periphery. This is probably due to uncorrected systematic errors that increase as a function of the distance from the image centre, error in the reference XYZ-stage itself near its maximum traveling distance, or variations in light intensity near the image periphery. However, even in this region, the system performance far exceeds the clinical requirement.

- The optimal metric performance of the VCM algorithms and the specific hardware used (without environmental effects) for this application can be summarized as follows: The precision (repeatability) has a standard deviation of 0.01 mm [1:30000] in X and Y, and 0.02 mm in Z [1:18000] while in terms of the system bias, the RMS of comparative measurements differences were 0.01 mm [1:18000] in X and Y, and 0.03 mm in Z [1:8000]. This optimal performance occurred for conditions where the field of view was 30cm x 30cm x 20cm, a base-to-standoff ratio of at least 0.7, and the target was at least 2.5 cm inside the field of view.

- To achieve those accuracy figures, images must be acquired at least 10 times and averaged, and the measurement cycle must be repeated at least 4 times. A single measurement may result in a maximum error of 0.03 mm under ideal conditions.

The VCM system as configured within the PPM has been implemented clinically to incorporate as many of the optimal conditions noted above as possible. However, since the achievable accuracy is much better than required accuracy (1 mm), cost

efficiency and performance can be optimized at the expense of the accuracy. For example, repeating and averaging of imaging and measurements will not be necessary, or desirable for this application. In addition, a fast target centroid method, applied to the binary image rather than the gray-scale image, will be sufficient. The best geometrical configuration may also be relaxed and, instead, a configuration to provide the best view of the treatment area can be employed.

3.2 Clinical Results

The PPM system was used for the clinical testing described herein. First, a prototype of the VCM software, but without the user interface and the special target/lighting system, was tested clinically at our facility for 8 months to assess its suitability for the application environment. After the system has been modified and completed with user interface and data analysis software, it was again clinically tested to verify and validate the performance of the complete system.

3.2.1. Initial Prototype Testing

During this period, 723 treatment fields on 23 patients were measured, providing 3703 available measurement points. The majority of the clinical measurements were performed on prostate, breast and lung patients. One patient was treated in a stereotactic frame designed for fractionated radiotherapy. All measurements were taken without feedback to the therapist. The data of figure 9 show the results of these measurements, displayed as a histogram showing the total displacement R_{dev} with respect to the initial setup of day one for each patient. The treatment/setup error for all measurements of these patients had a mean and mode of 5.5 mm and 4.0 mm respectively. Analysis reveals that in this small group of patients, in excess of 13.6% of the radiation dose was delivered with a total setup error of 10 mm or more.

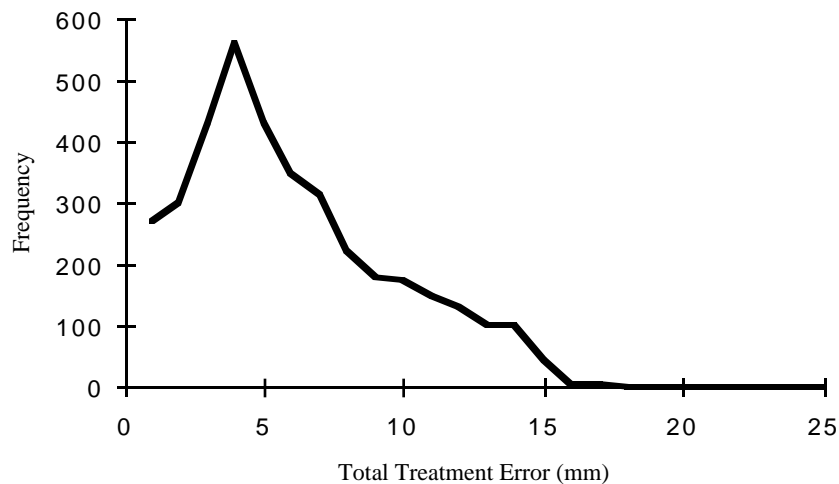


Figure 9: Histogram of movement of patients between treatments

The clinical use of the PPM system can best be illustrated anecdotally. The data of figure 10 portrays the application of the PPM system under clinical conditions requiring very precise patient positioning. In this case a patient with a brain lesion is treated in our relocatable stereotactic frame with fixed conformal fields. Markers were attached to the stereotactic frame and to the patient's forehead. The data are extracted from a total of 94 measurements taken over 7 fractions, where each data point is the average position for the specific fraction. In this figure, we present the setup/treatment accuracy of the patient and the stereotactic frame. The difference between these being the setup reproducibility of the patient within the stereotactic frame. As can be seen the stereotactic frame was relocated with a reproducibility of 1 mm., while the repositioning accuracy of the patient with respect to the treatment beam is better than 2.1 mm in all cases.

Figure 11 shows data acquired during the treatment of a four field prostate boost, representative of the application of the PPM system under clinical conditions where the precision would generally be considered less stringent than in the stereotactic case above. The data depicts the measured patient position for the first three fractions of a 10 fraction boost. The patient was treated with 9.5 cm x 8.0 cm AP-PA and 9.0 cm x 8.0 cm lateral fields. During treatment of the 2nd fraction, the patient experienced significant discomfort due to a full bladder. As can be seen both the Y (vertical) and Z (longitudinal) displacements increased from the moment of patient setup. When the third field was complete the placement error had become

significantly greater than would be normally accepted. At this point the patient was removed from the treatment couch, allowed to void, and then setup and treated for the fourth field which was well within acceptable limits.

Several problems arose with the initial application of the prototype VCM system in the clinical radiation therapy environment. Principal among these was the dependence upon lighting conditions, which strongly effected the ability of the VCM to recognize and extract the targets placed on the patient surface making it impossible to use the PPM system to measure the patient position during setup while the room lights were off and only the linac field light was visible. A second problem associated with lighting related to patients being treated with perspex immobilization shells. In this case, the VCM system occasionally interpreted reflections from the shell as targets. Both of these problems have been rectified with the introduction of the retroreflective targets, the 810 nm diode lasers and band pass filters on the camera lens,

The geometric restrictions of the radiation therapy room impose significant constraints on camera placement. In the ideal environment the camera would be placed above the patient on either side of the isocentre in the plane $Z = 0$, such that the angle subtended at the patient by the cameras would be greater than 38.6 degrees. There are several clinical and environmental constraints which make this impossible. Firstly, both cameras must view the patient in the vicinity of the isocentre, independent of gantry angle. This requires that the camera be moved forward away from the gantry. Another geometric problem arises from the fact that the patient surface is curved and the cameras have an oblique view of the patient. This restricts the area on the patient's surface which is suitable for target placement, since each target must be visible to both cameras. We have found that the cameras can be oriented within the room to provide an adequate view of the patient while only approximately maintaining optimal metric performance, i.e. an angle of greater than 38.6 degrees subtended at isocentre.

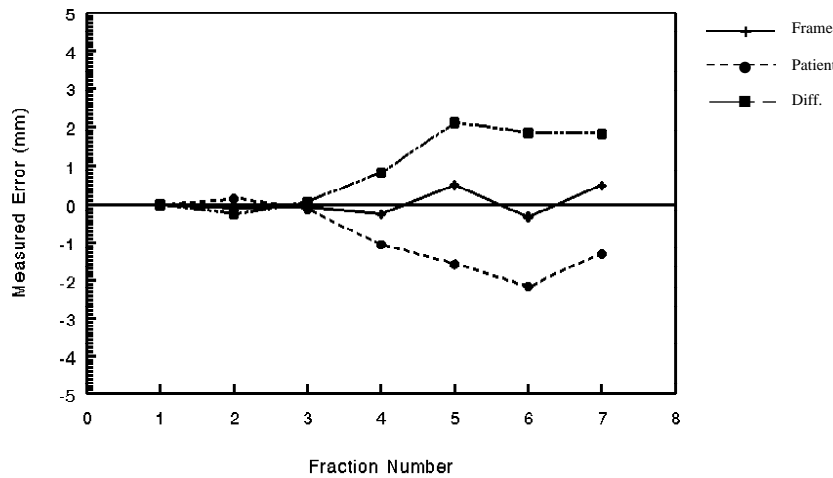


Figure 10: Positioning error for a patient with brain lesion

3.2.2. Complete System Testing

The redesigned version of the PPM software, with all the software and hardware components, have been tested in the clinical environment. The results of the repeatability and accuracy tests are shown below. The repeatability tests were carried out using four targets placed on a stable jig. The targets were measured 120 times at the same position and the standard deviation of the three position coordinates and the three orientation angles of the jig were computed. The results are:

$$\sigma_x = 0.02 \text{ mm}, \quad \sigma_y = 0.24 \text{ mm}, \quad \sigma_z = 0.22 \text{ mm}$$

$$\sigma_{\text{tilt}} = 0.033^\circ, \quad \sigma_{\text{roll}} = 0.087^\circ, \quad \sigma_{\text{yaw}} = 0.028^\circ$$

The accuracy was evaluated using a linear stage, for the position, and a rotating table, for the yaw angle (only this angle has been possible to evaluate by this set up). More than 10 measurements have been performed for each coordinate or angle and the deviation from the true values (assumed to be the external device reading) were computed.

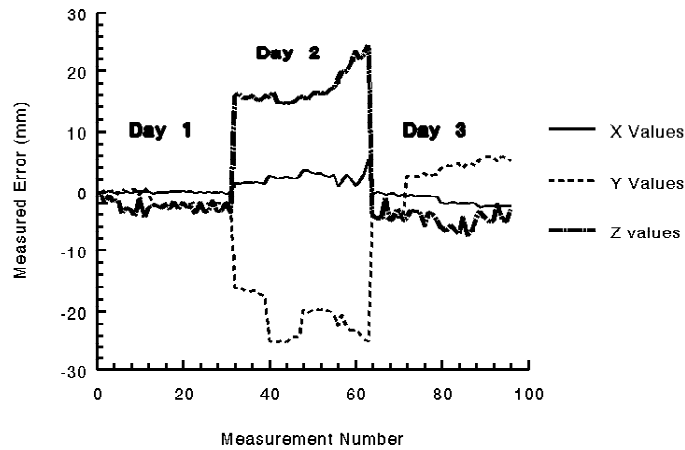


Figure 11: Positioning error for a patient with prostate cancer

The accuracy values at 99% confidence level are:

$$X_{acc.} = 0.05 \text{ mm}, \quad Y_{acc.} = 0.11 \text{ mm}, \quad Z_{acc.} = 0.21 \text{ mm}$$

$$yaw_{acc.} = 0.076^\circ$$

4. ANALYSIS AND DISCUSSION

Our clinical studies have shown that the system provides a clinically useful tool for patient position monitoring. The retroreflective target system has improved the performance of the system, making it immune to reflections and independent of ambient light conditions. The system is able to measure patient motion (target motion) within the range of specification, with an accuracy of better than 0.5 mm and with a frequency of greater than 0.2 Hz. The technologist has instantaneous feedback on patient position during treatment and during setup. It is expected that this feedback process should virtually eliminate large errors shown in figures 9 and 11.

The PPM system should be considered in comparison with real time portal imaging. These two measurement modalities are complementary to each other. They can easily be compared as follows:

- PPM can only measure the motion and position of the patient surface in space. Real time portal imaging on the other hand provides a direct image of the anatomy treated by the therapy beam.

- In contrast to real time portal imaging, PPM can measure patient position before delivering radiation to the patient.

- The PPM system has been demonstrated to have an accuracy and resolution in determining linear shifts (X,Y,Z) and rotation in all three planes which exceed the clinical requirement. Real time portal imaging on the other hand exhibits shortcomings in this area, particularly in terms of its insensitivity to motion parallel to the radiation beam axis and to determining rotations about any axis other than about the beam centre.

- The total time required to measure patient motion is of the same order for both systems. However, PPM acquires the information for determining the position in less than 0.2 seconds, such that there is little motion artifact in the reported position. Real time portal imaging provides an integrated measurement of position over a period of approximately 1 second.

- The results of the PPM measurement are interpreted and displayed directly to the user every 5 seconds. There is virtually no interpretation required by the user. Real time portal imaging on the other hand can provide updated images at the rate of about 1 Hz. The time required for the user to interpret the image is a function of the information and resolution to be retrieved. Automated interpretation algorithms are rapidly improving and can presently provide some registration information to the user in less than 30 seconds.

- The operation of the PPM system is independent of field size. In contrast, the usefulness of real time portal imaging is strongly dependent upon the size of the field irradiated.

It can be concluded from the above that the two methods of measuring patient position are in fact complementary. They measure different quantities and have correspondingly different strengths and weaknesses.

In conclusion we have developed a system capable of accurately and rapidly measuring patient position in Cartesian space before and during patient treatment. The physical parameters and resolution of the system have been well described under laboratory conditions. The full range of clinical applicability and limitation of this system is yet to be defined. The preliminary data presented herein suggest that this is an ideal tool for precise measurement of patient position for many different types of treatment and has the potential to be integrated into a patient record and verify system.

REFERENCES

1. Byhardt, R.W., Cox, J.D., Hornburg, A., Liermann G. "Weekly localization films and detection of field placement errors", *Int. J. Radiat. Onc. Biol. Phys.* **4**,881-887 (1978).
2. Kihlen B., Ruden B.I. "Reproducibility of field alignment in radiation therapy - A large-scale clinical experience", *Acta Oncologica* **28**,689-692 (1989).
3. Lam, W.C., Partowmah, M., Lee, D.J., Wharam, M.D., Lam, K.S. "On-line measurement of field placement errors in external beam radiotherapy", *BJR* **60**,361-367 (1987).
4. Rabinowitz, M.D., Broomberg, J., Goitein, M., McCarthy, K., Leong, J. "Accuracy of radiation field alignment in clinical practice", *Int. J. Radiat. Onc. Biol. Phys.* **11**, 1857-1867 (1985).
5. Rosenthal, S.A., Galvin, J.M., Goldwein, J.W., Smith, A.R., Blitzer, P.H. "Improved methods for determination of variability in patient positioning for radiation therapy using simulation and serial portal film measurements". *Int. J. Radiat. Onc. Biol. Phys.* **23**, 621-625 (1992).
6. Svensson, G.K. "Quality assurance in radiation therapy: physics efforts", *Int. J. Radiat. Onc. Biol. Phys.* **10**, 23-29, (1983).
7. Brahme, A. "Dosimetric precision requirements in radiation therapy", *Acta. Radiol. Onc.* **23**,379-391 (1984).
8. Goitein, M., Busse, J. "Immobilization error: Some theoretical considerations". *Radiology.* **117**,407-412 (1975).
9. McParland, B.J. "Uncertainty analysis of field placement error measurements using digital portal and simulation image correlations", *Med. Phys.* **20**, 679-685 (1993).
10. Webb, S., Nahum, A.E. "A model for calculating tumour control probability in radiotherapy including the effects of inhomogeneous distributions of dose and clonogenic cell density", *Phys. Med. Biol.* **38**, 653-666 (1993).
11. Bijhold, J., Gilhuijs, K.G.A., van Herk, M. "Automatic verification of radiation field shape using digital portal images" *Med. Phys.* **19**,1007-1014 (1992).
12. Dunscombe, P.B., Fox, K., Loose, S., Leszczynski, K. "The investigation and rectification of field placement errors in the delivery of complex head and neck fields", *Int. J. Radiat. Onc. Biol. Phys.* **26**,155-161 (1992).
13. Evans, P.M., Gildersleve, J.Q., Morton, E.J., Swindell, W., Coles, R., Ferraro, M., Rawlings C., Xiao, Z.R., Dyer, J. "Image comparison techniques for use with megavoltage imaging system", *BJR* **65**,701-709 (1992).
14. Ezz, A., Munro, P., Porter, A.T., Battista, J., Jaffray, D.A., Fenster, A., Osborne, S. "Daily monitoring and correction of radiation field placement using a video-based portal imaging system: a pilot study", *Int. J. Radiat. Onc. Biol. Phys.* **22**,159-165 (1991).
15. Gilhuijs, K.G.A., van Herk, M. "Automatic on-line inspection of patient set-up in radiation therapy using digital portal images". *Med. Phys.* **20**, 667-677 (1993).
16. Leszczynski, K.W., Shalev, S., Gluhchev, G. "Verification of radiotherapy treatments: computerized analysis of the size and shape of radiation fields", *Med. Phys.* **20**, 687-694 (1993).
17. Meertens, H., Bijhold J., Strackee, J. "A method for the measurement of field placement errors in digital portal images", *Phys. Med. Biol.* **35**, 299-323 (1990).
18. Reinstein, L.E., Sujatha P., Meek, A.G. "Assessment of geometric treatment accuracy using time-lapse display of electronic portal images", *Int. J. Radiat. Onc. Biol. Phys.* **22**, 1139-1146 (1992).
19. El-Hakim, S.F. "A hierarchical approach to stereo vision", *Photogrammetric Engineering and Remote Sensing.* **55**,443-448 (1989).
20. El-Hakim, S.F. and N.Pizzi, "Multicamera vision-based approach to flexible feature measurement for inspection and reverse engineering," *Optical Engineering*, **32**(9), 2201-2215 (1993).
21. Daher, R., McAdam, W., Pizey, G. "Implementation of a 3-D stereovision system for the production of customized orthotic accessories", in *Industrial Vision Metrology*, Proc. SPIE **1526**,90-95 (1991).
22. Brown, D.C. "Close-range camera calibration", *Photogrammetric Engineering*, **37**,855-866 (1971).
23. Karara H.M.(ed.). *The Handbook of Non-Topographic Photogrammetry*, 2nd ed., American Society of Photogrammetry and Remote Sensing, Falls Church, Virginia, USA (1989).
24. Leveson, N.G., "Software safety: why, what, and how", *Computing Surveys*, **18** (2), 125-163 (1986).
25. Eisenhart, C. "Realistic evaluation of the precision and accuracy of instrument calibration systems". *J. of Research of the National Bureau of Standards-C. Engineering and Instrumentation* **67C**,161-187, (1963).