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PHYSICS CONTRIBUTION

A RADIOGRAPHIC AND TOMOGRAPHIC IMAGING SYSTEM INTEGRATED INTO A MEDICAL LINEAR ACCELERATOR FOR LOCALIZATION OF BONE AND SOFT-TISSUE TARGETS

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Purpose: Dose escalation in conformal radiation therapy requires accurate field placement. Electronic portal imaging devices are used to verify field placement but are limited by the low subject contrast of bony anatomy at megavoltage (MV) energies, the large imaging dose, and the small size of the radiation fields. In this article, we describe the in-house modification of a medical linear accelerator to provide radiographic and tomographic localization of bone and soft-tissue targets in the reference frame of the accelerator. This system separates the verification of beam delivery (machine settings, field shaping) from patient and target localization.

Materials and Methods: A kilovoltage (kV) x-ray source is mounted on the drum assembly of an Elekta SL-20 medical linear accelerator, maintaining the same isocenter as the treatment beam with the central axis at 90° to the treatment beam axis. The x-ray tube is powered by a high-frequency generator and can be retracted to the drum-face. Two CCD-based fluoroscopic imaging systems are mounted on the accelerator to collect MV and kV radiographic images. The system is also capable of cone-beam tomographic imaging at both MV and kV energies. The gain stages of the two imaging systems have been modeled to assess imaging performance. The contrast-resolution of the kV and MV systems was measured using a contrast-detail (C-D) phantom. The dosimetric advantage of using the kV imaging system over the MV system for the detection of bone-like objects is quantified for a specific imaging geometry using a C-D phantom. Accurate guidance of the treatment beam requires registration of the imaging and treatment coordinate systems. The mechanical characteristics of the treatment and imaging gantries are examined to determine a localizing precision assuming an unambiguous object. MV and kV radiographs of patients receiving radiation therapy are acquired to demonstrate the radiographic performance of the system. The tomographic performance is demonstrated on phantoms using both the MV and the kV imaging system, and the visibility of soft-tissue targets is assessed.

Results and Discussion: Characterization of the gains in the two systems demonstrates that the MV system is x-ray quantum noise-limited at very low spatial frequencies; this is not the case for the kV system. The estimates of gain used in the model are validated by measurements of the total gain in each system. Contrast-detail measurements demonstrate that the MV system is capable of detecting subject contrasts of less than 0.1% (at 6 and 18 MV). A comparison of the kV and MV contrast-detail performance indicates that equivalent bony object detection can be achieved with the kV system at significantly lower doses (factors of 40 and 90 lower than for 6 and 18 MV, respectively). The tomographic performance of the system is promising; soft-tissue visibility is demonstrated at relatively low imaging doses (3 cGy) using four laboratory rats.

Conclusions: We have integrated a kV radiographic and tomographic imaging system with a medical linear accelerator to allow localization of bone and soft-tissue structures in the reference frame of the accelerator. Modeling and experiments have demonstrated the feasibility of acquiring high-quality radiographic and tomographic images at acceptable imaging doses. Full integration of the kV and MV imaging systems with the treatment machine will allow on-line radiographic and tomographic guidance of field placement. © **1999 Elsevier Science Inc.**

Portal imaging, Conebeam computed tomography, Kilovoltage, Megavoltage.

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INTRODUCTION

The escalation of tumor dose in radiotherapy promises increased probability of disease control (1). The proximity of the target to normal tissues makes dose escalation challenging. Safe pursuit of these higher doses has spurred the development of conformal therapy techniques. These techniques attempt to conform the radiation field to a well-specified target volume. The success of the conformal therapy approach requires (i) accurate characterization of the uncertainties in field placement to create reasonable margins for use in the planning process, and (ii) verification of the treatment delivery.

Numerous studies of field placement error have been reported in the literature over the past 10 years (2–5). These studies have demonstrated that on-line megavoltage portal imaging is useful in measuring field placement uncertainty. Many of these studies also report on the poor quality of the portal images and how the quality of the images hampers the detection of field placement errors (5).

The primary reason for the poor quality of the megavoltage portal images is the intrinsically low subject contrast of bony anatomy at megavoltage energies (6). The dramatic drop in contrast with increasing x-ray energy requires that the noise introduced by the imaging system be extremely low. In the case of conventional verification/localization film, the low contrasts are masked by film noise and the fixed display contrast of the film (7). Fluoroscopic systems suffer from poor light collection and electronics noise in the light sensor and readout electronics (8). While many investigators continue to investigate methods of reducing noise in these systems, we have taken a more direct approach: to increase the subject contrast by using a kilovoltage (kV) x-ray source to localize the placement of the field.

Using a kV x-ray source to determine field placement is not a novel concept. Johns *et al.* (9) report the design of a ^{60}Co treatment unit with a kV x-ray tube in the head to guide field placement. Investigators have reported on the addition of an x-ray tube to the head of an accelerator to generate kV radiographs of the patient in treatment position (10–12). Other investigators have demonstrated that a low-energy x-ray beam can be produced by a low atomic number, transmission target in the head of the accelerator (13). In 1992, Suit *et al.* (14) urged the community to develop better techniques for verifying field placement, proposing the addition of a kV source to the accelerator. In a previous report, we described the development of a prototype digital imaging system for kV localization on a medical linear accelerator (15).

Beyond the obvious advantage of improved image quality for portal imaging, the development of an on-line kV imaging system can provide other advantages. The dose delivered in a kV exposure is significantly lower than that required for a megavoltage image. This reduction in dose allows more frequent imaging with open-fields, and the acquisition of images at non-treatment gantry angles, which report on the characteristics and performance of an MV imaging system and a kV imaging system that have been integrated with a medical linear accelerator for radiographic localization. The dosimetric advantage of kV radiographs for localization is examined and the potential for tomographic guidance is demonstrated with the system using cone-beam computed tomography (CB-CT). The combined system consisting of the medical accelerator and the two on-line imaging systems (kV and MV) is referred to as the dual-beam system (DBS).

MATERIALS AND METHODS

Dual-beam system

Overview. Photographs of the DBS are shown in Fig. 1. Two fluoroscopic imager assemblies are attached to the accelerator; one detects the megavoltage (MV) treatment beam, the other detects the kV beam projected at 90° to the treatment beam axis. An Elekta SL-20 (Elekta Oncology Systems, Crawley, UK) linear accelerator forms the basis of the system. This accelerator is computer controlled and produces 6 and 18 MV photon beams. Field shaping is performed with an 80 leaf collimator. The SL-20 is a drum-based accelerator, making the installation of a retractable kV x-ray source relatively straightforward (Fig. 1d).

A 600,000 heat unit (HU) x-ray tube (Eureka Rad-92 in Varian Sapphire housing, 0.6 and 1.2 mm focal spots, 12.5° rotating anode; Varian X-ray Tube Products, Arlington Heights, IL, USA) with a manual collimation system has been installed on the SL-20 with the focal spot located 100 cm from the machine isocenter. The tube is supported by 3 hardened steel shafts (nominal 2.5” diameter) that retract into the accelerator’s drum structure (Fig. 1d). The shafts are supported by three bearing assemblies set in an aluminum mount. The aluminum mount is attached to the face of the accelerator drum structure at the position corresponding to a 90° gantry angle. Adjustments are built into the assembly for focal spot alignment. The total weight of the x-ray tube and mechanical components is approximately 90 kg. Under this load, and at full extension, the predicted deflection in focal spot position was approximately 0.2 cm. The x-ray tube is powered by a 40 kW high-frequency, radiographic generator (Innerscan Inc., Chicago, IL, USA). The exposure-to-charge ratio in air for this tube is measured to be 5.7 mR/mAs at isocenter (100 cm) for a 90 kVp spectrum. The tube spectrum has been modeled based on measured first and second HVLs of aluminum (Al) (HVL_1 = 4.0 mm; HVL_2 = 9.5 mm) (16).

The MV radiographic imager assembly is shown schematically in Fig. 2 and is similar in design to that reported by Atari *et al.* (17). Briefly, a phosphor screen (Gd_2O_2S:Tb, Fuji HR-16 [back], 165 mg/cm^2) mounted on a 1.5-mm stainless steel plate is used for primary x-ray detection. The light emitted by the phosphor screen is reflected by three front-surface mirrors and focussed by a lens (f/0.95 operating at f/1.4, 50 mm) on to a 512 × 512 back-thinned

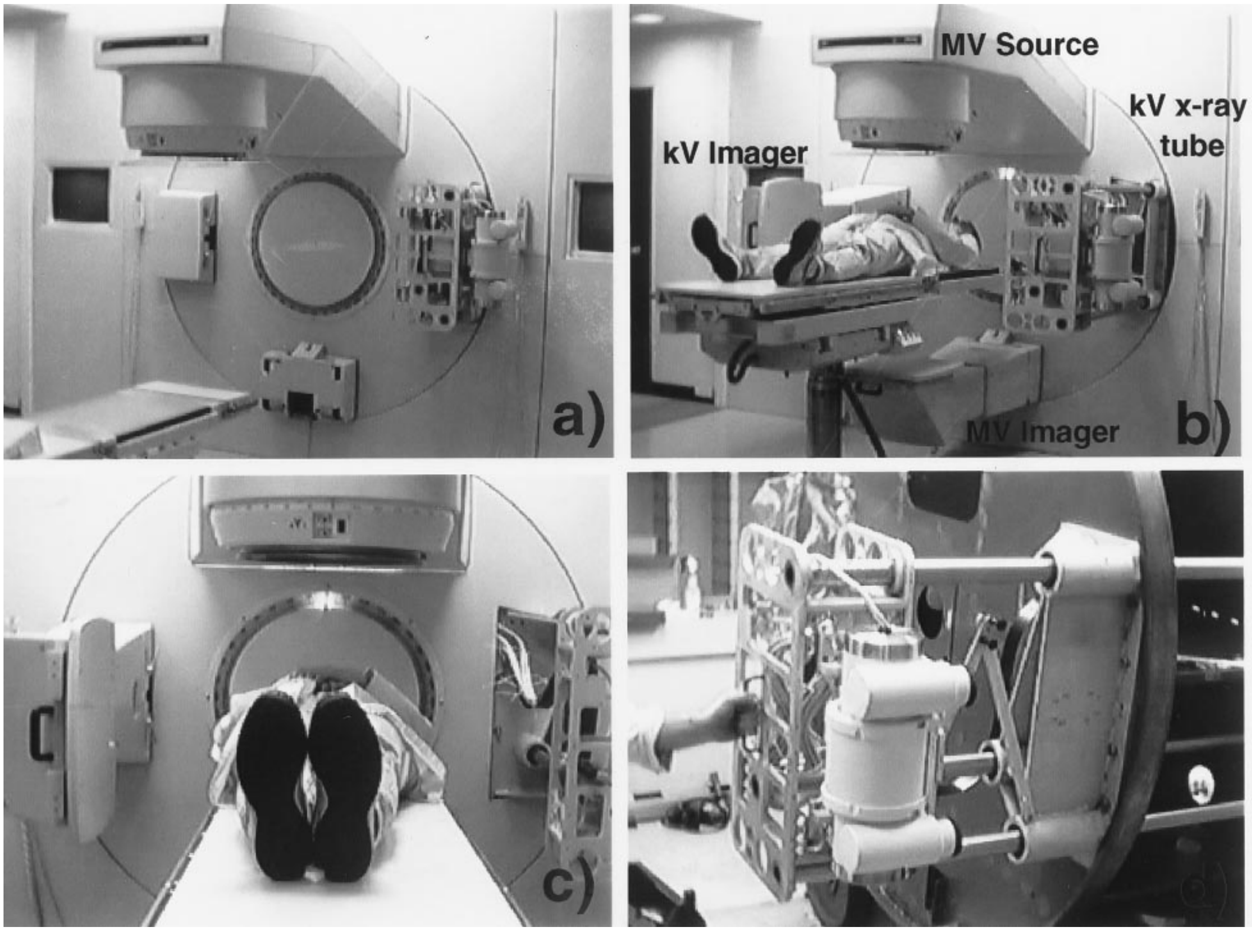

Fig. 1. Photographs of the dual-beam system. The dual-beam system was constructed on a Elekta SL-20 medical linear accelerator. The design allows normal operation of the accelerator with the imagers removed and the kV x-ray tube retracted (a). The imager can be easily attached for kV localization. The kV tube is extended manually to reach the source plane (b). The focal spot of the x-ray tube is at 100 cm SAD providing the same imaging geometry as that of the MV source. This provides a large aperture for conebeam CT with the system (c). The drum structure of the SL-20 simplifies the addition of the retractable arm of the x-ray tube. Three hardened-steel shafts support the x-ray tube (d).

Imaging Technologies [SITE] Inc., Beaverton, OR, USA). Back-thinning provides a quantum efficiency of 78% at 550 nm (peak in the Gd_2O_2S:Tb spectrum). The CCD sensor is housed in a Photometrics CH-250 camera housing (Photometrics Inc., Tucson, AZ, USA). The sensor is cooled to −40°C to minimize the dark current collected in the sensor during read-out. The sensor has a full-well capacity of 280,000 electrons, corresponding to the full-range of the 12-bit digitizer. A shutter is installed to minimize the effects of light leaks in the imager housing. The camera is read out under computer control at 500 kpixels/s. The long read-out period (0.5 s) of the camera prevents operation in a fluoroscopic mode.

The kV imager assembly is identical to the MV assembly with the exception of one less mirror in the optical path and the replacement of the 1.5 mm stainless steel plate with a thin Al plate (1/32") for support of the phosphor screen. The kV imager assembly is a refined design of the original MV assembly; the third mirror has been eliminated through modifications to the mounting assembly. The presence of of the MV system. In the case of the megavoltage detector, the stainless steel plate increases the interaction probability for the MV X rays and reduces the signal generated by scattered X rays exiting the patient. The imager housing (Fig. 1b) for the MV system has been modified to allow retraction by ~34 cm. The kV imager housing is non-retractable and is attached and removed before and after an imaging procedure.

An in-depth characterization of the imaging performance of the kV and MV radiographic systems is on-going and is not the focus of this article.

Control system. The DBS is slaved to the operation of the linear accelerator. A personal computer (Pentium-based) controls the x-ray generator and the acquisition of images and also monitors the gantry angle reported by the accelerator. A schematic of the control system is shown in Fig. 3. The generator is controlled via an RS-232 port, allowing full-control of the generator's parameters and control of the exposure. The cameras are read via two ISA control cards in the PC. The gantry angle is monitored via the accelerator's

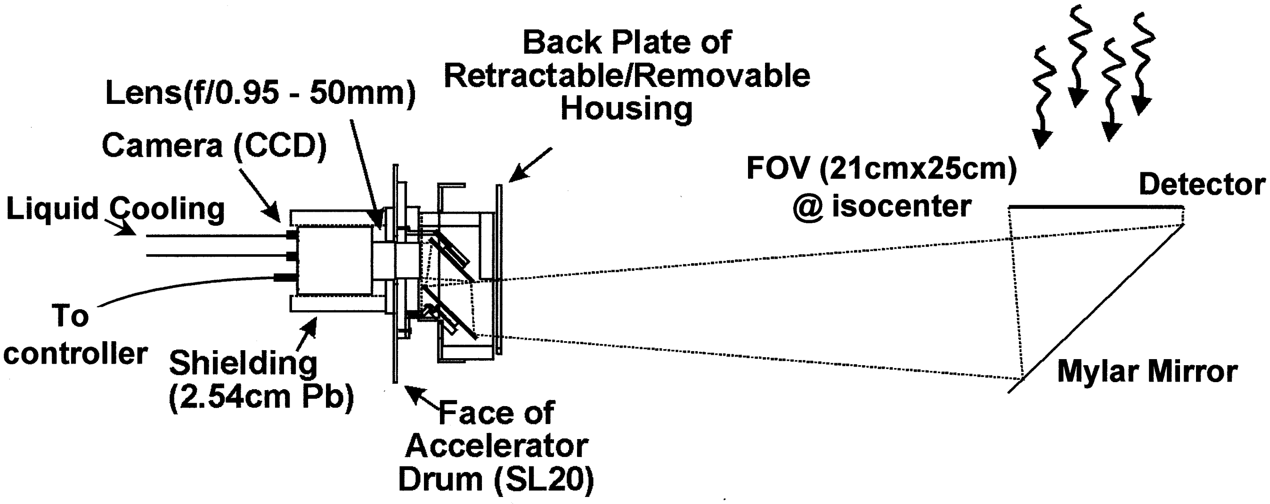

Fig. 2. Schematic of the imaging systems used to acquire the MV and kV radiographs. Both systems are based on a cooled CCD camera for collecting light emitted by a 165 mg/cm^2 Gd_2O_2S:Tb screen. The kV and MV systems are identical with the exception of an additional mirror in the MV assembly and the use of different metal plates. The kV assembly is a newer generation of the MV design. The housing for the MV assembly has been altered to permit retraction of the screen and large mirror allowing the couch to lower to the floor completely.

to 12 bits by an analog-to-digital board in the control PC. The gantry angle can be determined to within ±0.1 degrees with this mechanism.

System performance

Mechanical characteristics of the system. The DBS was constructed to measure the patient position with respect to the reference frame of the medical linear accelerator. The localization accuracy and precision that can be achieved with this system will depend on the mechanical stability or

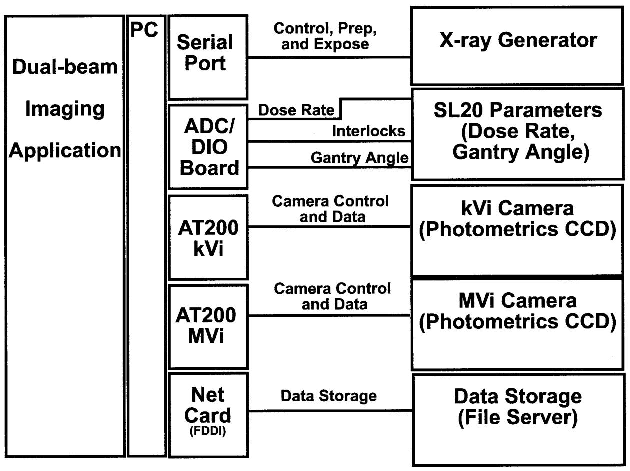

Fig. 3. The kV and MV imagers are controlled via a PC. This figure shows the mechanisms used to control and monitor the different components of the system. The gantry angle of the accelerator is monitored via an analog voltage line provided through the accelerator's interface. The images acquired by the system are stored in a distributed database. The same system is used to acquire the many projections used for the conebeam

rigidity of the system's components. To estimate the precision with which high-contrast objects can be located, measurements were made of the range of flex for each of the four components of the DBS as the gantry was rotated through 360°. These four components include: the kV x-ray source, the MV x-ray source, the MV imager assembly, and kV imager assembly.

The magnitude of accelerator drum wobble and eccentricity was determined by directing a small (<1 mm diameter) laser beam onto the face of the drum. The gantry was rotated through 360° and the path of the laser spot on the drum was plotted. The laser was then adjusted to the geometric centroid of the path and the process was repeated. After a few iterations, the maximum extent of the path was determined to be the magnitude of eccentricity at the face of the drum. The measurements were repeated with a rigid mechanical extension securely fastened to the face of the drum. The extension moved the plane of measurement 1 m out from the face of the drum (isocenter lies 1.25 m out from the face of the drum). This experiment, in conjunction with the results from the face of the gantry, allowed us to approximate the magnitude of accelerator (drum) wobble at the plane of the x-ray source orbit (isocenter).

The run-out of the gantry (motion of the entire drum assembly along the axis of rotation) was also measured. This was done with the use of a dial gauge attached to the front face of the accelerator (near the axis of rotation). The tip of the dial gauge was placed in contact with the end of the accelerator's couch. The tip of the gauge was positioned on the gantry's axis of rotation through an iterative procedure similar to that described in the previous paragraph. The gantry was then rotated through 360° and the relative move-

Table 1. Mechanical characteristics of the DBS components

	Component			
Range (mm)	MV source	MV imager	kV source	kV imager
Tangential	1.5	4.0	4.0	1.5
Axial	1.1	1.1	1.1	1.1
Radial	2.0	5.0	5.0	1.5
Drum	0.6	0.6	0.6	0.6

The measured range of motion of each of the system's four components. To establish a conservative estimate for localizing precision, it is assumed that each of these motions are completely independent. The ability to correct for these motions depends on their reproducibility as a function of gantry angle; preliminary investigations suggest that such corrections could be applied reliably.

recorded. Care was taken to verify that the couch was rigid and immobile as the gantry rotated.

To estimate the range of flex of the two x-ray source assemblies and the two imager assemblies, a solid state laser was mounted rigidly on the face of the drum and its beam (after passing through a cylindrical lens) was directed onto each of the four system components in-turn. As the gantry was rotated, the intersection of the line projected by the laser with the component was recorded on a small strip of paper attached securely to the surface of the component. The range of relative movement between the projected laser beam and the structure was taken to be due to flex in the component. The laser (and projected line) was then rotated and the measurement was repeated in the orthogonal dimension (Table 1). It is important to note that the results of this measurement present a combination of the component's flex and any shearing of the assembly. In addition, they do not determine the path of the component but provide a reasonable estimate for the range of the non-rigid motion. A more complete characterization of component flex is on-going, in which the movements of each component are characterized in three separate dimensions (radial, longitudinal, and angular) as a function of the nominal gantry angle (18).

With these estimates of component flex, the geometric precision of the DBS for three-dimensional localization was estimated assuming an orthogonal-pair imaging application, which uses either the MV or kV system. The calculations were performed assuming the three-dimensional point of interest is near machine isocenter. Using the uncertainties in the viewing geometry introduced by the mechanical non-rigidity of the system, it was possible to calculate the localizing precision using the following method. At a given gantry angle, the range of angular motion of the source (kV or MV) and the range of angular motion of its corresponding detector assembly define a truncated sector in the isocenter plane. In an orthogonal view, the same ranges of motion define a second truncated sector in the isocenter plane. The extent of the area represented by the intersection of these two truncated sectors is taken as a measure of the

Table 2. Stages in imaging system and their corresponding gains

		Detector and X ray energy	
Stage in imaging system	Gain (quanta exiting stage/quanta entering stage)	Megavoltage imager (6 MV)	Kilovoltage imager (90 kVp)
1	g_1 (interacting X ray/ incident X ray)	0.025	0.67
2	g_2 (optical quanta produced/interacting X ray)	1.96×10^4	3.05×10^3
3	g_3 (fraction of optical quanta escaping screen)	0.7	0.7
4	g_4 (fraction of optical quanta reflected by mirrors while *en route* to lens)	0.83	0.88
5	g_5 (fraction of optical quanta collected by lens)	1.36×10^{-4}	1.42×10^{-4}
6	g_6 (quantum efficiency of CCD)	0.78	0.78

Listing of the gain stages in the imaging system. Scattering stages have not been included in this analysis. The values used for each stage were determined directly or taken from the literature.

or away from isocenter) of the components have negligible effects on the shape and location of the intersection defined here, and, therefore, have been ignored in this analysis. These calculations are performed to indicate the worst-case condition in which no corrections for mechanical flex have been applied.

Radiographic imager performance: kilovoltage and megavoltage

The kV and MV imaging systems operate on the same principle: the interaction of an x-ray within the phosphor screen (or metal plate/phosphor screen) produces light, which is collected by a high-speed lens and focussed onto an efficient, low-noise CCD. Each stage in the two systems has been characterized to examine the transfer of signal through the imaging chain. Table 2 lists the six stages included in the analysis. The gain or interaction probability for each stage has been estimated or taken from the literature. From these values, it can be determined whether the two systems are operating near their theoretical limit of performance, as dictated by the noise associated with x-ray quanta used to form the image. This analysis is restricted to low-spatial frequencies and does not consider the small amount of additive noise introduced by the CCD camera.

The gain estimates for the first stage (g_1) were determined using Monte Carlo techniques (19) for the MV detector at 6 MV (20) and using a spectral model (90 kVp) for the kilovoltage detector (16). The conversion of x-ray energy to light was determined from the Monte Carlo generated (MV) or analytical (kV) estimates of energy deposition in the phosphor layer. For the kV detector, K-fluorescence and

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