FLAT-PANEL CONE-BEAM COMPUTED TOMOGRAPHY FOR IMAGE-GUIDED RADIATION THERAPY

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Purpose: Geometric uncertainties in the process of radiation planning and delivery constrain dose escalation and induce normal tissue complications. An imaging system has been developed to generate high-resolution, soft-tissue images of the patient at the time of treatment for the purpose of guiding therapy and reducing such uncertainties. The performance of the imaging system is evaluated and the application to image-guided radiation therapy is discussed.

Methods and Materials: A kilovoltage imaging system capable of radiography, fluoroscopy, and cone-beam computed tomography (CT) has been integrated with a medical linear accelerator. Kilovoltage X-rays are generated by a conventional X-ray tube mounted on a retracted arm at 90° to the treatment source. A 41 \times 41 \text{ cm}^2 flat-panel X-ray detector is mounted opposite the kV tube. The entire imaging system operates under computer control, with a single application providing calibration, image acquisition, processing, and cone-beam CT reconstruction. Cone-beam CT imaging involves acquiring multiple kV radiographs as the gantry rotates through 360° of rotation. A filtered back-projection algorithm is employed to reconstruct the volumetric images.

Conclusions: A kV cone-beam CT imaging system based on a large-area, flat-panel detector has been successfully adapted to a medical linear accelerator. The system is capable of producing images of soft tissue with excellent spatial resolution at acceptable imaging doses. Integration of this technology with the medical accelerator will result in an ideal platform for high-precision, image-guided radiation therapy.

Radiotherapy, Cone-beam computed tomography, Linear accelerator, Image guidance, Flat-panel detectors.
illustrate the many challenges in reducing these uncertainties below current levels. Investigations of internal organ motion (9–20) have demonstrated that, for many sites, substantial reductions in geometric uncertainty require the visualization of internal structures in the reference frame of the treatment machine. The development of volumetric imaging systems for online image guidance has been a major focus of research in the past 5 years.

Many investigators have examined the use of the treatment beam to perform megavoltage (MV) computed tomography (CT) of the patient in treatment position. This was first proposed in 1983 by Swindell et al. (21) and has been extended to a cone-beam implementation by Mosleh-Shirazi et al. (22) Other investigators have employed various two-dimensional (2D) detectors to perform MV cone-beam CT on conventional accelerators (23–26). In 1987, Brahme et al. (27) proposed the development of MV CT based on the 50-MV scanning photon beam of the racetrack microtron; this approach offers elevated contrast due to the increasing pair-production cross-section at X-ray energies over 1.022 MeV. More recently, Ruchala et al. have reported on the development of a helical MV CT scanner in conjunction with a tomotherapy unit (28). Although many investigators have been evaluating MV CT for radiotherapy verification, the only system to reach routine clinical use is the one developed by Nakagawa et al. (29). In their procedure, a pretreatment MV CT slice is used to verify the patient setup for stereotactic radiosurgery of the lung. Although utilization of the MV source for imaging seems to offer an elegant solution in terms of imaging and delivery with the same source, it faces the enormous challenge posed by the poor detection efficiency of X-ray detectors in the MV energy range (30). The low efficiency results in poor signal-to-noise performance for clinically acceptable doses (<10 cGy). Furthermore, the increased radiation transport in the X-ray detector reduces the spatial resolution that can be expected at these energies.

Introducing existing imaging technologies into the therapy setting is another alternative. The clear value of integrated imaging and delivery compelled Uematsu et al. to install a conventional CT scanner and a conventional simulation unit in the radiotherapy suite (31). This approach offers volumetric imaging and real-time fluoroscopy in the treatment room. Reference between the imaging and delivery systems is maintained through a single, pivoting table that can dock to each system. The system has been employed in a variety of treatment sites, demonstrating the many advantages of integrated imaging and delivery (32–34).

It has been our approach to integrate a kilovoltage (kV) X-ray source and large-area flat-panel detector on a medical linear accelerator for fluoroscopy, radiography, and cone-beam volumetric CT (8, 35–38). The cone-beam method allows a volumetric CT image to be reconstructed from data collected during a single rotation of the gantry. This approach offers the combined functionality that has been identified by Uematsu et al. (31) in a single integrated imaging and delivery system. Geometric cross-calibration of the imaging and delivery systems allows volumetric images to be collected in the reference frame of the delivery system (8, 38). In this form, a highly integrated model can be envisioned in which the control system of the treatment unit orchestrates the interplay of the imaging and delivery components on a single machine. Such an approach offers the flexibility to employ a treatment-procedure-specific imaging strategy—whether real-time fluoroscopy, radiography, cone-beam CT, or an appropriate combination of all three. Integration will allow image-guided procedures to be performed within the tight time constraints found in the radiation therapy setting.

In previous publications, the feasibility of cone-beam CT based image-guidance has been explored using a kV imaging system based upon a phosphor screen optically coupled to a charge-coupled device (CCD). The drawbacks of the CCD-based approach have been identified (39) and a new approach based upon large-area flat-panel detectors has been proposed (40). The adaptation of this technology to a medical linear accelerator and an evaluation of its performance is the subject of this paper.

**METHODS AND MATERIALS**

**Imaging system**

A kV imaging system capable of radiography, fluoroscopy, and cone-beam CT has been integrated with a medical linear accelerator (Elekta, SL-20, Elekta Oncology Systems, Atlanta, GA). Kilovoltage X-rays are generated by a conventional X-ray tube (Eureka Rad-92, Varian Sapphire Housing, see Table 1) which has been mounted on a retractable arm that extends from the accelerator’s drum structure (Fig. 1a). The focal spot of the tube is located at 90° to the MV source and 100 cm (± 0.2 cm) from the rotation axis of the accelerator, sharing a common axis of rotation with the MV treatment source. The X-ray tube is powered by a 45-kW high-frequency X-ray generator that operates under computer control. A 41 × 41 cm² flat-panel X-ray detector (Fig. 1b; RID 1640, PerkinElmer Optoelectronics, Wiesbaden, Germany) is mounted opposite the kV tube at a gantry position of −90° and a nominal detector-to-focal spot distance of 155 cm. The detector has many features that make it ideally suited to cone-beam CT including a high X-ray cross-section and efficient, distortion-free read-out. Such detectors were initially developed for radiography and fluoroscopy applications, and these areas continue to drive improvements in their performance. The characteristics of the detector used in this system are listed in Table 1. The entire imaging system operates under computer control with X-ray exposure and detector readout synchronized via software running under the Windows NT operating system (Fig. 2a). Calibration, image acquisition, processing, and cone-beam CT reconstruction execute on a personal computer equipped with a 500-MHz Pentium Xeon processor and 1 GB of RAM.

**Imaging geometry**

Accurate cone-beam CT reconstruction from X-ray radiographs requires that the projection geometry for each
radiograph is known at the time of reconstruction. For systems that are well-described by a simple rotation, the projection geometry can be calculated from the angular increment provided the source, axis of rotation, and detector positions are well known for at least one projection. The system under investigation here cannot be described by a simple rotation. The substantial mass of the X-ray tube and detector combined with their distance from the drum face (125 cm) result in substantial flex of these components during gantry rotation.

The calibration procedure is similar to that reported by Fahrig and Holdsworth (41) and begins by placing a small steel ball bearing (0.8 cm diameter) at the nominal treatment isocenter of the treatment gantry as represented by the in-room laser system. X-ray projections are taken of the bearing as the gantry rotates clockwise through 360°. The centroid of the bearing in each image is determined automatically via software and recorded. The overall displacement of the projected bearing location with respect to the detector is measured to verify that the bearing is reasonably placed (within 0.2 cm of the nominal axis of rotation). From this analysis, a calibration map of centroid position vs. gantry angle is established. Due to the presence of hysteresis effects in the motion of the gantry, the calibration procedure is performed for a specific start angle (−180°) and direction (clockwise).

### Table 1. Characteristics of imaging system components

<table>
<thead>
<tr>
<th>Component</th>
<th>Specification</th>
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</thead>
<tbody>
<tr>
<td>Linear Accelerator</td>
<td></td>
</tr>
<tr>
<td>Manufacturer</td>
<td>Elekta Oncology Systems</td>
</tr>
<tr>
<td>Model</td>
<td>SL-20, Sl/MLCi</td>
</tr>
<tr>
<td>Mass</td>
<td>∼5 ton</td>
</tr>
<tr>
<td>Range of rotation</td>
<td>± 185°</td>
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<tr>
<td>Maximum angular velocity</td>
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<tr>
<td>kV X-ray Source</td>
<td></td>
</tr>
<tr>
<td>Manufacturer</td>
<td>Varian Associates</td>
</tr>
<tr>
<td>Model</td>
<td>RAD-92, Sapphire</td>
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<tr>
<td>Mass (tube, housing, collimator)</td>
<td>44.6 kg</td>
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<tr>
<td>Anode heat capacity</td>
<td>600 kHU</td>
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<tr>
<td>Target angle</td>
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<tr>
<td>Focal spot size (nominal)</td>
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<td>Inherent filtration</td>
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<td>Additional filtration</td>
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<td>Generator</td>
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<td>Manufacturer</td>
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<td>Model</td>
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</tr>
<tr>
<td>Mass</td>
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<td>X-ray converter</td>
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<tr>
<td>Detector size</td>
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<td>Pixel pitch</td>
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<td>Fill factor</td>
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<td>Photodiode capacitance</td>
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<tr>
<td>Photodiode bias voltage (V)</td>
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<tr>
<td>Photodiode charge capacity</td>
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</tr>
<tr>
<td>Charge transfer time</td>
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</tr>
<tr>
<td>ASIC charge capacity</td>
<td>5 pC</td>
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<tr>
<td>Maximum frame rate</td>
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<tr>
<td>Detector dynamic range</td>
<td>&gt;83 dB</td>
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<td>Image lag</td>
<td>&lt;3%</td>
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<tr>
<td>Digitization</td>
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</table>
The measured centroid positions are taken to be the location of the central axis on the detector face (sometimes referred to as the “piercing point”). The back-projection process of the reconstruction code has been modified to compensate for the variations in central axis position from projection to projection through a simple look-up table. These corrections are equivalent to applying small in-plane translations in detector position at the time of back-projection and do not fully account for axial displacements of the X-ray focal spot.

**Imaging procedure**

Table 2 provides a summary of the imaging geometry and acquisition/reconstruction parameters for the imaging sys-

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**Fig. 2. (a) Screen capture representation of the Windows NT–based application developed to acquire, process, reconstruct, and store projection and cone-beam CT datasets. Fields for patient information, treatment data, and image datasets are integrated with the clinical database. Processes for image acquisition and processing are similar to those previously described (40), and reconstruction is performed using cone-beam filtered back-projection (42). The host workstation used in this work was a 500 MHz Pentium Xeon PC with 1 GB of RAM. (b) Timing diagram illustrating synchronization of imager readout, gantry rotation, and X-ray exposure. The period between image frames is constant throughout acquisition ($T_{\text{frame}} = 570$ ms). Gantry rotation initiates before the first projection and is measured by means of a calibrated potentiometer circuit. Gantry angle is sampled twice at each projection—immediately before and immediately after delivery of the X-ray exposure—and averaged to yield the angle of the projection view.**
In the current system, X-ray generation and image acquisition operate independently of gantry motion. The gantry rotation speed is selected to provide a reasonable angular velocity and detector readout rate were selected such that velocity was maximized while not requiring extremely precise timing tolerances in the synchronization of detector readout and X-ray exposure. The timing diagram for the imaging procedure is illustrated in Fig. 2b. In this approach, the image acquisition interval is set by the frame rate of the detector. The X-ray exposure immediately follows image storage, with the gantry angle sampled immediately before and following each exposure. The average of the two samples indexes the previously acquired projection calibration files and generates a geometric description of the projection. This image acquisition cycle continues until the gantry has completed 360° of rotation. X-ray exposures (120 kVp; 25 mA; 25 ms in this investigation) were reproducible within ~3%, with slight variations in tube output normalized according to the detector signal measured in a region of the field that was not occluded in any projections. To achieve fast data transfer and storage, all projections for a single acquisition (330 images nominally) were stored in host memory (660 MB allocated for storage of projection data).

Immediately before acquisition, a set of 20 frames is read from the detector in the absence of X-rays to characterize the offset present in the detector signal. An additional set of dark frames is acquired in conjunction with a set of open-field exposures at the same technique as employed during the cone-beam CT acquisition. Based on these images, each projection of the cone-beam CT data set is corrected for pixel-to-pixel offset and gain variations. Defective pixels in the sensor have been previously identified, and the pixel values at these locations are corrected using a 3 × 3 median filter. The defective pixels are detected semiautomatically in a separate series of tests performed on the detector. Briefly, pixels are excluded automatically according to criteria such as pixel gain, dark current, and dark current stability. This typically identifies 99% of the defective pixels with the remaining 1% identified by visual inspection of defect-corrected images. For the detector used in these investigations, approximately 5,000 of the 1,048,576 pixels were identified as defective.

The algorithm employed to reconstruct the cone-beam CT images from the projections is based on that reported by Feldkamp et al. (42). Briefly, the projections are weighted and filtered in one dimension. The resulting images are interpolated at four times resolution and stored for sampling during back-projection. As stated above, the geometric calibration for each image modifies the pixel look-up and back-projection vectors to accommodate flex in the mechan-
ical assembly and variations in interprojection angular increment. The influence of flex on the reconstructed images can be evaluated by comparing images reconstructed with and without the flex corrections applied.

**Phantom studies**

Overall performance of the system was evaluated using two phantoms: (i) an anthropomorphic head phantom, and (ii) a contrast phantom oriented coronally.

The anthropomorphic phantom (The Phantom Laboratory, Salem, NY) consists of the head and torso of a human skeleton embedded in plastic. With the head of the phantom suspended off the end of the treatment couch, 321 projections (120 kVp, 25 mA, 0.025 s) were acquired over 183 s as the gantry rotated through 360°. The in-air exposure (in the absence of phantom) delivered to the isocenter for this technique is 1.2 R. Four image sets were reconstructed from the projection data including a cubic volume (341 × 341 × 341 at 0.075 cm voxel pitch) and three orthogonal slices (transverse, coronal, and sagittal) at full resolution (1024 × 1024 at 0.025 cm voxel pitch).

The contrast phantom is shown schematically in Fig. 3a and consists of twelve 2.9-cm-diameter inserts placed in a 16.7-cm-diameter water bath 5 cm in depth. The inserts and their nominal CT numbers are listed in Fig. 3a (43). The phantom was placed at the isocenter of the accelerator gantry on a radio-translucent support (Fig. 3b). The X-ray technique was identical to that employed for the head phantom (see above). Three hundred thirty projections were acquired over 360° requiring 188 s. Four adjacent coronal slices were reconstructed at full resolution (1024 × 1024 at 0.025 cm voxel pitch) and averaged to produce a 0.1-cm-thick coronal slice. Images of the contrast phantom were also acquired on a conventional CT scanner (Philips SR-
7000, Philips Medical Systems, Atlanta, GA) using a non-helical technique (120 kVp, 200 mA, 1 s, Standard Head Filter, 0.1 cm slice thickness, 0.1 cm table index). The technique delivered a 4.5 R exposure to isocenter (measured in the absence of the phantom). The entire length of the phantom was scanned in 180 s, generating 180 slices. A $25 \times 25$ cm$^2$ field-of-view (FOV) was reconstructed at a resolution of 0.049 cm ($512 \times 512$).

**RESULTS**

**Geometric nonidealities**

The graphs in Fig. 4 describe the measured geometric nonidealities in the present system. Figure 4a presents an analysis of the measured gantry angle during three different cone-beam CT acquisitions. The angular velocity of the gantry was found to vary by a factor of 2 over the roughly 180 s required to travel through $360^\circ$. Given that the projections are sampled at a fixed time interval (0.570 s), a higher density of projections is acquired near $0^\circ$ than at $\pm 180^\circ$. The angular increment varied from $0.7^\circ$ to $1.6^\circ$ over the course of the acquisition. No weighting corrections were applied to accommodate these variations.

In addition to the variations in angular increment, the system also suffers from variability in the geometric relationship between the kV focal spot and the flat-panel detector. The graph in Fig. 4b illustrates the influence of gantry angle on the projected isocenter at the detector plane. Assuming that the focal spot travels on a circular trajectory that is well described by gantry angle, the observed displacement in projected isocenter can be attributed to flex in the detector arm or gross motion of the entire gantry with respect to the treatment room. The plot in Fig. 5b suggests that both types of motion are occurring in this case. The deviations in the u-coordinate are indicative of flex in the arm, whereas deviations in the v-coordinate are indicative of translation of the gantry along its axis during rotation (i.e., a corkscrew motion). Overall, the magnitude of this motion is less than 0.2 cm in both dimensions. As listed in Table 2, the location of the projected isocenter is within $1.1 - 1.2$ pixels of the mean $(u, v)$ location on the detector ($517.6, 496.3$), where $(512, 512)$ denotes the center of the $1024 \times 1024$ detector matrix. The reproducibility of the observed flex has been evaluated through repeated (5) scans of the ball bearing over a 4-h period. The maximum standard deviation in the projected isocenter location $(u, v)$ for each gantry angle was determined to be 0.4 and 0.2 pixels (1 pixel equals 0.25 mm at isocenter), and the average standard deviation over $360^\circ$ was 0.1 and 0.05 pixel. The measured flex data acquired most proximally to an imaging procedure was employed in the reconstruction. The variations in angular increment and observed flex are taken into consideration during reconstruction by adjusting the back-projection to be consistent with the geometry of acquisition.

**Phantom studies**

Cone-beam CT reconstructions of the anthropomorphic head phantom are shown in Fig. 5. The full FOV reconstruction (Fig. 5a) was performed at one-third resolution (0.075 cm voxel pitch) to illustrate the large FOV achieved with a single rotation of the gantry in the cone-beam ap-
proach. The large FOV and high spatial resolution of the cone-beam system are illustrated in Figs. 5b–d. Each of these images was reconstructed from the same cone-beam CT data set. The transverse image (Fig. 5b) is the conventional presentation of CT data. Figures 5c and 5d illustrate the fully volumetric nature of the cone-beam CT data, showing excellent spatial resolution in all three dimensions. Trabecular bone structure is clearly visible in the vertebral column (Figs. 5e and 5f). Overall, the images exhibit a uniform quality of contrast and spatial resolution performance throughout the reconstruction without severe influence from noise or image artifacts, and the potential for high-precision radiotherapy of the head and neck becomes evident. Consistent with previous investigation of the modulation transfer function (40), the system provides submillimeter spatial resolution (~0.7 mm full-width at half-maximum of the line-spread function). A slight vertical streak artifact can be seen in the sagittal cut, the origin of which is under investigation.

Figure 6 illustrates the importance of correcting for flex and gantry displacement during reconstruction. Cone-beam reconstructions of the anthropomorphic head phantom were performed with the flex correction scheme disabled and compared with those reconstructed with the flex correction active. The influence of the small corrections (<0.2 cm) is very clear when the corrected and uncorrected reconstructions are compared. The nonrigidity is manifest in three ways: (1) an overall reduction in the visibility of detail (as evident in the anterior aspect of the hard palate in Figs. 6d and 6e); (2) the introduction of misregistration artifacts (as seen by a double bone edge along the anterior aspect of the posterior cranial fossa, directly adjacent to the mastoid air cells in Figs. 6a and 6b); and (3) the introduction of streak artifacts (horizontal streaks through the mastoid air cells in Fig. 6b). The streak artifacts are expected to become more pronounced for higher contrast objects, possibly obscuring surrounding low-contrast structures of interest. These images illustrate the importance of minimizing geometric inconsistencies between acquisition and reconstruction. Every effort should be made to eliminate geometric nonidealities or guarantee their reproducibility.

The imaging results for the coronal contrast phantom are presented in Fig. 7. The cone-beam CT image (Fig. 7a) was reconstructed at a voxel pitch of 0.025 cm with four adjacent slices averaged to effect a 0.1 cm slice in the coronal plane. Overall, the image contains significant spatial detail.

Fig. 5. Cone-beam CT images of the anthropomorphic head phantom. (a) Surface rendering of the volume data identifying axial, coronal, and sagittal planes and illustrating the uniform volumetric spatial resolution of the data. The image in (a) was reconstructed using a (26.5 × 26.5 × 26.5) cm³ field-of-view with 0.075 cm voxels (i.e., ~3 times the nominal voxel pitch of 0.025 cm). (b) Axial, (c) coronal, and (d) sagittal cuts through the volume data reconstructed at full resolution (1024 × 1024 voxels at 0.025 cm voxel pitch). The data illustrate the uniform quality and quantity of image data acquired in a cone-beam CT acquisition. The high spatial resolution is illustrated further by magnifying the bony anatomy surrounding the brainstem (e and f).
as well as soft-tissue contrast. The lowest readily detectable contrast insert is breast at 47 Hounsfield units (HU). An image of the same phantom acquired on the conventional CT scanner is presented in Fig. 7b. A direct comparison of spatial resolution and contrast detection is not possible due to differences in the voxel size (factor of 4 in voxel volume) and total dose used to form the two images. However, it is interesting to note that the two scans were acquired in the same amount of time (180 s) and the conventional CT scanner was operated at its limit of spatial resolution in the axial dimension (slice thickness of 0.1 cm). This difference in axial resolution is clearly visible in Figs. 7c and 7d. For the cone-beam CT data, resolution is comparable to that in the transaxial plane (~0.05 cm). For the conventional CT image, on the other hand, z-resolution is limited by the slice thickness (in this case, 0.1 cm). A low-contrast streak artifact in the x-dimension can be seen around the higher contrast bone inserts. This type of artifact is typically seen in the transverse plane for CT systems with a geometric miscalibration and results from inconsistencies in the back-projection of opposing views. The presentation of such artifacts in the coronal plane indicates an axial misregistration among projections. The artifact is most likely due to subtle variations in the axial displacement of the gantry between acquisitions. Previous investigations have demonstrated hysteresis effects for this motion (i.e., the axial position of the gantry depends on both angle and direction of rotation). Further study of the stability of these corrections is required.

Overall, the cone-beam CT image demonstrates the sys-
tem’s capacity for high spatial resolution and soft-tissue imaging. In addition to these features, a comparison of cone-beam and conventional CT highlights the lack of any obvious geometric distortions in the cone-beam CT image.

DISCUSSION

A kV cone-beam CT system has been adapted to the gantry of a medical linear accelerator for image-guided radiation therapy. The results demonstrate that the system is capable of imaging human anatomy with excellent spatial resolution and soft-tissue sensitivity. Prior investigations have demonstrated that flat-panel detectors are suitable for cone-beam CT (37, 40, 44–48). The results reported here demonstrate that they can be applied under nonideal geometric conditions. The magnitude of the geometric nonidealities observed in these investigations is not representative of what would be observed on all accelerator gantries. Furthermore, in-field service adjustments can easily be performed to minimize the observed corkscrew motion of the gantry. The variations in angular velocity seem to be reasonably well accommodated by the reconstruction code, and refinements that better accommodate nonequangular views are currently being implemented and tested.

One outstanding area of concern is in the increased presence of X-ray scatter in the cone-beam CT geometry. Scattered X-rays reaching the detector can lead to cupping artifacts and to a reduction in contrast-to-noise performance. Recent investigations have demonstrated that there are several ways of addressing these issues (47). The first approach is to minimize the amount of X-ray scatter reaching the detector while maintaining a high detector efficiency. Our results suggest that a source-to-detector distance between 150 and 160 cm is optimal for the system reported here under conditions of high X-ray scatter, such as would be found in imaging the pelvis. The lack of obvious scatter artifacts in the
images presented here supports this assertion. With the system geometry optimized, other methods of knowledgeably managing X-ray scatter can be pursued, including (i) proper selection of the voxel size during reconstruction and (ii) increasing the imaging dose. These choices require procedure-specific information, such as the required spatial resolution for therapy guidance, the dimensions of the imaged anatomy, and the clinical constraints on imaging dose (48).

It is the overall objective of these investigations to develop a system that permits the radiation oncologist to confidently deliver highly conformal radiation dose distributions in the human body with a level of conformation more in keeping with radiosurgical practice than conventional radiation therapy. Given the exquisite detail visible in the images presented here, it seems clear that such a system would begin to reduce the distinction between these two disciplines, not only allowing dose escalation but also challenging existing fractionation dogma. A great deal of work remains in the development of this technology into the form necessary for high-precision image-guided radiation therapy; however, there appear to be no technological roadblocks to this development. Given the results presented above, there are several issues that remain to be addressed if the overall objective is to be considered achievable, including image FOV, acquisition and reconstruction time, patient motion, cross-calibration of imaging and delivery components, and integration of the imaging system with the treatment control system.

In the system described above, the reconstruction FOV is limited to a cylinder 26.5 cm in length and 26.5 cm in diameter. This limitation is a result of the requirement that the entire extent of the imaged object remain in the detector FOV for all projections. Cho et al. (49) have developed a method to increase the reconstruction FOV of small detectors by offsetting the detector with respect to the central axis. This approach requires the collection of projections over a full 360°. An offset scheme with 7.6 cm of overlap at the detector would provide a 48-cm reconstruction FOV—comparable to that of a conventional CT scanner. While there is certainly the potential for larger detectors (>41 cm), the detector employed in this work represents the maximum size achievable with current flat-panel detector fabrication facilities. Therefore, the development of the offset detector scheme is currently of critical importance for sites below the head and neck.

In the current system, the acquisition time for a complete cone-beam data set is approximately 180 s. Improvements in system synchronization would allow these acquisition times to be reduced by a factor of 2 (i.e., 90 s), wherein a hardware circuit provides synchronized firing of the X-ray tube, and the detector is operated at its maximum readout rate of 0.285 s/frame. Faster frame rates (30 frames/s) have been reported for similar detectors (50), and such frame rates would allow 330 projections to be acquired in under 15 s. At this rate, the maximum angular velocity of the gantry (1 revolution per minute) becomes the limiting factor. This restriction is attributed to the International Electromechanical Commission (Geneva, Switzerland) recommendations with respect to the motion of medical devices and may need to be reconsidered in light of the advantages of faster image acquisition rates. Relatively slow acquisition rates will result in higher probability of motion artifacts in the reconstructed volume. Artifacts introduced by breathing and peristalsis are two areas that require further investigation, and these artifacts may restrict use in these areas until faster systems can be developed.

The clinical application of cone-beam CT for image-guided therapy in an online setting requires images to be available within a few moments of acquisition. The images presented in this paper were reconstructed on a 500-MHz Pentium Xeon processor equipped with 1 GB of RAM requiring from a few minutes for a single high-resolution transverse slice to 21 h for a 341 × 341 × 341 cube. Very little effort has been invested in accelerating these reconstruction times, because high-speed reconstruction engines are commercially available and can be readily adapted for cone-beam CT reconstruction of the data presented here (private communication, H. Bruning, Ph.D., TeraRecon Inc., San Mateo, CA; private communication, R. Brauner, Mercury Computing Inc., Chelmsford, MA). Such systems are capable of reconstructing 256° volumes from 300 projections (1024 × 1024) in under 60 s (private communication, H. Bruning). Given this level of performance, the imaging data would be available within 1 min of completion of acquisition; furthermore, given the potential for reconstruction during the acquisition process, the imaging data would be available within seconds of completion of acquisition.

A significant difference between conventional CT and cone-beam CT in the context of image-guided radiation therapy is that cone-beam CT data can be referenced directly to the treatment machine coordinates without the use of skin marks, body frame, or calibrated couch movements. The use of cone-beam CT for guiding therapy requires the development of a calibration scheme to cross-calibrate the imaging and treatment components of the accelerator. In this approach, every voxel in the reconstructed image can be located relative to the MV source and various field-shaping devices. An interface is envisioned in which the graphical representations of these components can be presented relative to the recently acquired cone-beam CT images. Of course, if this system were capable of calculating dose it would be of great interest to visualize the dose with respect to the imaged anatomy. Thus, it is becoming clear that a reasonable interface to the accelerator may contain many of the features currently found in a 3D treatment planning system. The cone-beam system also has the significant advantage of high-quality fluoroscopy and radiography in the treatment room, permitting simplified radiographic verification of patient positioning as well as fluoroscopic monitoring during the delivery process—the latter being of particular importance with the increased use of small treatment fields as found in step-and-shoot or dynamic IMRT.

Finally, an area of critical importance for clinical implementation of this technology is in the seamless integration of the imaging system within the control system of the linear accelerator. It is clear that clumsy imaging procedures
will not succeed in today’s busy treatment centers, and only through the development of a highly integrated imaging and delivery system will this technology begin to make clinical inroads. An unfortunate case in point can be found in the very slow acceptance of portal imaging technology. Although this technology has been available in one form or another for many years, it has yet to be adopted in the mainstream. This is a consequence of the lack of appropriate tools for image interpretation and failure to fully integrate these systems into a clinically acceptable practice. It is hoped that we can learn from this experience and not have the technology developed here suffer a similar fate.

The clinical application of this technology also needs to be carefully explored. This exploration should be stratified by treatment sites with the relative merits of image-guided radiation therapy explored in each context. While it is clear that many treatment sites would gain through increased delivery precision, radiation therapy of the prostate has been of particular interest in our investigations, as it illustrates many of the challenges found in other sites. The proximity of rectal wall and prostate often requires overlapping avoidance and planning target volumes, significantly restricting dose escalation. Soft-tissue imaging at the time of delivery will allow online planning and may offer nonoverlapping rectal and planning target volumes. In this scenario, hypofractionation becomes feasible without risking an increase in rectal toxicity. Other sites of application include head and neck for visualization of soft-tissue anatomy in the neck, visualization of muscle compartments in the treatment of sarcoma, and image-guided treatment of metastases (such as in the liver), to name a few. In addition to the active role this technology may play in influencing the precision of delivery, there is also a very significant passive role that should not be understated. For the first time in radiation therapy practice, the opportunity for accurate documentation of the fraction-by-fraction delivery of the radiation dose will be feasible. The importance of this added information on our understanding of dose response in tumors and the dose-volume relationships in disease and normal structure requires no further elaboration.

CONCLUSION

Overall, the results of these investigations support the hypothesis that cone-beam CT based on large-area, flat-panel detectors is an excellent candidate for image-guided high-precision radiation therapy. Previous investigations of the fundamentals of flat-panel performance and their application to cone-beam CT have demonstrated that such an approach can provide the performance necessary for soft-tissue imaging in the treatment geometry. The results reported here demonstrate that this technology can be successfully adapted to a conventional medical linear accelerator accommodating the presence of geometric nonidealities. Although it is clear that the technology is capable of producing volumetric images of a patient in treatment position, it may not be immediately clear how these images will be employed in directing therapy.

We propose the development of a new generation of radiation therapy treatment machines capable of integrated imaging and delivery, providing a means for seamless target identification, real-time monitoring, and efficient delivery. Such a system is based upon kV cone-beam CT imaging and conventional cone-beam radiation delivery. In this approach, the advantages inherent in the clinical flexibility of the conventional gantry structure are further heightened through the capacity for low-dose soft-tissue imaging and fluoroscopic monitoring.

REFERENCES

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