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Dynamic intensity-weighted region of interest imaging for conebeam CT

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Abstract

BACKGROUND—Patient dose from image guidance in radiotherapy is small compared to the treatment dose. However, the imaging beam is untargeted and deposits dose equally in tumor and healthy tissues. It is desirable to minimize imaging dose while maintaining efficacy.

OBJECTIVE—Image guidance typically does not require full image quality throughout the patient. Dynamic filtration of the kV beam allows local control of CT image noise for high quality around the target volume and lower quality elsewhere, with substantial dose sparing and reduced scatter fluence on the detector.

METHODS—The dynamic Intensity-Weighted Region of Interest (dIWROI) technique spatially varies beam intensity during acquisition with copper filter collimation. Fluence is reduced by 95% under the filters with the aperture conformed dynamically to the ROI during cone-beam CT scanning. Preprocessing to account for physical effects of the collimator before reconstruction is described.

RESULTS—Reconstructions show image quality comparable to a standard scan in the ROI, with higher noise and streak artifacts in the outer region but still adequate quality for patient localization. Monte Carlo modeling shows dose reduction by 10–15% in the ROI due to reduced scatter, and up to 75% outside.

CONCLUSIONS—The presented technique offers a method to reduce imaging dose by accepting increased image noise outside the ROI, while maintaining full image quality inside the ROI.

Keywords

Region of interest imaging; image guidance; IGRT; cone-beam CT

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1. Background

In addition to its traditional role in diagnostics, CT is widely utilized as a guidance tool in therapeutic procedures. This has been driven, in a large part, by advances in flat-panel x-ray detector technologies that have enabled the rapid development of flat-panel cone-beam CT (CBCT). Guidance systems typically consist of a flat panel detector mounted opposite a kilovoltage x-ray tube, often on a C-arm in interventional radiology and image-guided surgery, or mounted directly to a linear accelerator for image-guided radiation therapy (IGRT). These systems have enabled new therapies and made existing therapies more accurate and safer for the patient. However, CT imaging involves additional x-ray exposure to the patient, and we must consider both the “As low as reasonably achievable” (ALARA) principle [1] and the AAPM Task Group 75 report [2], which states that it is no longer safe to consider imaging dose negligible and recommends “that strategies for reducing the imaging dose and volume of exposed anatomy be pursued wherever possible, even when they require developing new image acquisition and reconstruction techniques”.

Dose to the patient from CBCT imaging can be reduced by altering the x-ray fluence, beam energy or the number of views acquired. Most dose reduction strategies currently use a reduced fluence approach. Often this is done by reducing the tube current which can be done on a view-by-view basis to account for differences in patient thickness from different angles [3, 4]. It can also be done uniformly across all views, further reducing dose but resulting in noisier projection data. There has been work on statistically driven iterative techniques for denoising the data prior to reconstruction [5–7] or controlling noise during reconstruction [8–10]. Another approach to limiting the patient exposure in which there has recently been great progress is to reduce the number of angular samples, utilizing few-view reconstruction [11–13]. However, these dose reduction methods are global in nature, in that they are applied generally during acquisition without consideration of the requirements for the specific imaging task.

An alternate approach is to image only the region of interest (ROI) by restricting the x-ray field to only cover this region. This is often possible in guidance applications because, unlike diagnostics, the object of interest and its location have been previously determined and the imaging is being performed to confirm the location or locate an interventional device relative to it. With traditional reconstruction techniques, such as FDK [14], restricting illumination to only the ROI can result in truncation artifacts in the image. We previously reported a technique to overcome such limitations by using partially transmitting filters to substantially decrease the x-ray exposure to the regions outside the ROI while maintaining sufficient data for traditional reconstruction techniques [15, 16]. The concept is illustrated schematically in Fig. 1 where an interior ROI is illuminated with the full source fluence, and the surrounding region with a reduced fluence. In that work static filters were utilized, resulting in a cylindrical ROI centered on the gantry rotation axis.

2. Objective

Here we extend the intensity-weighted region of interest (IWROI) method to handle arbitrary shaped and positioned ROIs, by introducing a dynamic collimation device and

necessary data correction strategies for reconstruction with FDK. This technique has been named dynamic IWROI, or dIWROI, imaging [17–19].

For reconstruction the imaging physics in CT is often simplified to a model including only the exponential attenuation of the x-rays as they pass through object to be measured. This can be written as

$$I = I_0 e^{\left(-\int \mu(\vec{r}) dl\right)} \quad (1)$$

where I_0 is the incident beam intensity and I is the measured beam intensity at the detector plane. The local, linear attenuation coefficient is $\mu(\vec{r})$ which also represents the CT image. This model neglects non-linear aspects of the imaging physics, such as the energy dependence of μ and scatter, which will be discussed further in the following sections.

However, in this form the solution for $\mu(\vec{r})$ has been derived, and for the circular cone-beam CT case FDK is the most commonly used algorithm.

The model above is typically simplified a step further, such that the sinogram data (p) represents the log ratio of I_0 to I . In the form below the parameters (u, v, θ) are used for the cone-beam geometry such that each projection is a 2D image parameterized with u and v , and projections are taken from many angles, θ .

$$p(u, v; \theta) = \ln \left(\frac{I_0(u, v; \theta)}{I(u, v; \theta)} \right) = \int \mu(\vec{r}) dl \quad (2)$$

The intensity weighting technique varies the intensity of the x-ray beam with position. In this framework that is represented as a scaling factor on the intensity, $w(u, v, \theta)$. This scales both I and I_0 in Equation (1), and drops out in the log ratio in Equation (2). Thus, the intensity weighting does not impact the value of the sinogram data or by extension the values in the reconstructed image. Rather, the effect of the weighting is seen in the noise properties. The emission, attenuation and detection of the x-rays are all stochastic processes and reducing the beam intensity effectively reduces the number of measurements, thus increasing the noise in the data. The weighting factor can then be viewed as a local control on the noise level of the projection data. For FDK the propagation of the noise from the projection data is well localized, in a practical if not theoretical sense, meaning that the intensity weighting can be used to locally control the noise properties of the reconstructed image. Thus the weighting can be used to reduce the imaging dose to regions of the patient where higher noise levels can be tolerated for the given imaging task.

3. Methods

3.1. Imaging system

Imaging was carried out with the On-Board Imager (OBI) on a Trilogy linear accelerator (Varian Medical Systems, Palo Alto, CA). This standard clinical system can acquire CBCT datasets during gantry rotation, a full one-minute rotation typically resulting in roughly 640 projection views. The 30 by 40 cm detector panel has an effective pixel pitch of 0.388 mm. Generator settings for all presented results were 120 kVp, 80 mA and 13 ms per projection.

3.2. Dynamic collimation

The dynamic ROI imaging techniques presented in this work require dynamically restricting or modulating the intensity of the imaging beam upstream of the patient during the CT acquisition. The kV source on Varian linear accelerators is equipped with lead collimator blades which can define a rectangular aperture of arbitrary size and position anywhere within the detector area. However, the blades cannot be moved while the kV beam is on, so do not themselves provide a mechanism to achieve dynamic collimation nor can the blade material be altered to allow for different levels of partial transmission. Accordingly, we have designed an accessory dynamic kV collimator that can be attached to the source housing downstream of the blades, in place of the bowtie filter as shown in Fig. 2.

The intensity-weighting collimator blades were 3.2 mm thick copper, each mounted on 4 slide cars moving on low profile, low friction linear guide rails. The guide rails were mounted to an acrylic base machined to a tight surface flatness tolerance to prevent any binding in the blade motion. The base was mounted with 9.5 cm aluminum stand-offs to an aluminum mounting plate, designed to fit in place of the standard bowtie filter. The blades were driven by linear actuators (L16, Firgelli Technologies, Victoria, British Columbia) which have an unloaded peak speed of 20 mm/s and a nominal positional accuracy of 0.4 mm. With the collimator plane 251.6 mm from the focal spot the magnification factor was 3.975 resulting in a nominal isocenter plane speed of 79.5 mm/s and accuracy of 1.6 mm. For true conformal (i.e., ROI without intensity weighting) imaging requiring radiopaque blades, removable covers consisting of 3.2 mm lead backed by 0.8 mm steel were added.

The on-board control of the collimator blades utilized an Overo series computer-on-module (COM) (Gumstix Inc, Portola Valley, CA) with a TI OMAP 3503 600 MHz ARM Cortex A8 processor, 512 MB RAM and a 4 GB SD card for storage. The gantry rotation was measured using an inclinometer (US Digital, Vancouver, Washington) whose signal was accumulated by an Arduino Duemilanove microcontroller board with the signal lines attached to the interrupt pins of its processor, ensuring that no inclinometer pulses were missed. Serial-over-USB communication was used between the Arduino and Overo modules. The linear actuators were controlled by independent dedicated motor controllers (Pololu Corporation, Las Vegas, NV) providing closed loop feedback control.

3.3. System modeling/controller tuning

The motor controllers use a three term feedback control method commonly referred to as proportional-integral-derivative (PID) control. Good system performance depends on setting

the coefficients of the three PID terms appropriately. The process of determining these is known as tuning the controller.

In order to perform the tuning we performed a separate external position measurement using transmissive linear strip and optical encoder modules (LIN-200-6-N and EM1-0-200-N, US Digital). The encoder modules were attached to the collimator blades, while the transmissive strip was clamped between bars of aluminum and acrylic attached to the base plate. To ensure proper alignment with the blade motion, the strip was supported on custom made fine-threaded brass standoffs and locked in place between adjusting nuts. To explore possible gravitational effects on performance and tuning under controlled conditions, the system was mounted on a rotary stage so the entire collimator assembly could be rotated about a horizontal axis with the blades in a vertical plane. Otherwise the motor loads were the same as those experienced during imaging.

3.3.1. System model—In order to develop the mathematical system model for controller tuning the motors were driven in an open loop mode. The encoders were read out by a computer running Labview with a NI-7340 motion control card and nuDrive amplifier (National Instruments) which also monitored the drive signal on an analog input. The motors were manually driven through several arbitrary wave forms and their position recorded. A data set containing the motor input, the position and the time were fit to a transfer function model with 2 zeros and 4 poles. The best fit model was

$$P(s) = \frac{0.01306 + 0.2196s^{-1} - 0.2318s^{-2}}{1 - 1.927s^{-1} + 0.6694s^{-2} + 0.4457s^{-3} - 0.1882s^{-4}} \quad (3)$$

This model was then tested against a dataset from an independent run showing that the model predicted position agrees well with the measured position, as shown in Fig. 3. The slight model overshoot in the peaks and troughs is likely due to the limited sampling frequency of the input data.

3.3.2. Calibration—In order to calibrate the blade positioning to the imager coordinate system, the collimator was mounted to the kV source and each blade was separately stepped through various positions. At each step a radiograph was acquired. The set of radiographs was then processed to determine blade edge position. Specifically, for each image the intensity histogram was generated, which was bimodal with one peak corresponding to low intensities under the collimator blade and another to high intensities in the open field. The midpoint between these two peaks was taken as the edge crossing intensity threshold. The blade edge position was determined by fitting a spline function to the central image column, and identifying the point at which the spline crossed the edge threshold, as illustrated in Fig. 4. This procedure was automated in Matlab; it communicated with the collimator directly, prompted the user when to acquire a radiograph and automatically processed the images when all acquisitions were completed.

3.3.3. System accuracy—After installation and calibration the system performance was evaluated with regard to accuracy under static conditions with stationary gantry and single blade setpoint, and dynamic conditions with rotating gantry and continuous blade motion. The static test was performed driving the blades to a set of fixed set points with the imager in a lateral radiograph position. The collimator was given ample settling time to reach its final position. At each set point a single image was acquired and the blade edges were automatically located using the process described above. For both blades, with set points over the full image range, the mean measured error was 2.75 pixels which equates to 0.7 mm at the isocenter.

The dynamic test was performed using a blade trajectory for a 5×7 cm elliptical ROI located 3.5 cm off the axis of rotation. Three consecutive full rotation CBCT scans were taken with nothing in the field of view. The blade edge positions were automatically extracted as described above.

In the dynamic case, with rotating gantry and continuous collimator motion, the observed errors were larger, ranging from −10 to 20 pixels. However, while the blade position may deviate from the ideal trajectory there is high repeatability between scans. Using the deviation from the mean observed position, rather than the planned position, we found that the scan to scan variation was less than ±1.75 pixels. Much of the observed error in the dynamic case is likely due to predictable system changes during rotation, such as gravitational sag of the arms that carry the kV tube and the detector panel. For this application, these errors are tolerable since they are repeatable. It is the alignment of the blade edges between the object scan and the air scan used in the reconstruction that is most critical, and conformal imaging will utilize a non-zero margin around the ROI in any case. A calibration procedure could also be implemented which uses measured variation in imager component positions with gantry angle to correct the blade trajectories, eliminating this source of variability.

3.4. Imaging experiments

Two phantoms were used for real data studies. The first was the head section of the RANDO Man phantom (The Phantom Laboratory, Salem, NY) which is a standard anthropomorphic phantom used for dosimetry in radiation therapy. It consists of human skeleton cast in a soft tissue equivalent urethane plastic. The natural skeleton provides good, high contrast, detailed structure, however the phantom lacks any low contrast structures relevant for soft tissue discrimination in a clinical setting. The second phantom was designed to provide soft tissue information. It was a common supermarket roasting chicken, with a bag of internal organs and ordinary bread cube stuffing in the thoracic cavity.

Prior to the CBCT imaging studies, both phantoms were scanned on the Philips Brilliance Big Bore 16-slice helical CT scanner in the radiation oncology clinic, shown in Fig. 5. The images were then imported into the Pinnacle³ treatment planning system (Philips Electronics, Amsterdam, Netherlands), and ROI contours were drawn. This process of acquiring a planning scan and marking relevant regions of the anatomy in the treatment planning system is a standard part of the patient treatment workflow. We then extracted the

ROI contours from Pinnacle³ and projected them onto the detector plane to determine the apertures required for conformal illumination.

3.5. Conformal ROI imaging

With lead added to the copper dynamic collimator blades, we were able to perform the type of dynamic x-ray field shaping required for true conformal ROI imaging with the clinical imaging system. Conformal ROI results are shown here merely as a point of reference for the proposed intensity weighted technique, and use the chord-based backprojection filtration (BPF) reconstruction techniques originally proposed by [20] as they are of the greatest familiarity to the authors. The ROIs for this technique were chosen as peripheral regions with illumination satisfying published data sufficiency conditions [21, 22]. Under these conditions it has been shown that the ROI can be accurately reconstructed.

3.6. Data corrections

Practically, spatially varying the intensity of an x-ray beam for a projection taken within a fraction of a second is non-trivial. The approach taken here is to use the dynamic collimation device described above with copper filters to partially occlude the beam. This device imposes several restrictions on the intensity weighting capabilities. The weighting function effectively becomes binary with the open aperture being “unfiltered” and the remainder of the view being “filtered”. The aperture is constrained to be rectangular and aligned with the detector axes and the change in the aperture from view to view is limited by the speed of the collimator.

In order to calculate the sinogram data (p) in Equation (2) above, the incident (I_0) beam intensity must be known as well. In order to effectively capture the structure caused by the heel effect of the tube anode, the inverse square law intensity fall off across the flat detector and spatially varying detector sensitivity, I_0 is usually measured directly as a separate scan with nothing in the field of view. For the weighting factor to drop out properly it must be identical in I and I_0 , meaning the blade edge positions must match exactly between the two datasets. This is mechanically challenging so additional software corrections were developed. Additionally, the hardening of the beam spectrum by the copper filters must also be accounted for. An overview of the complete processing chain is shown in Fig. 6.

Processing I_0 —Noise in the I_0 scan contributes to noise in the reconstructed image. Typically this is a small contribution as the intensity of the image is high and the noise can be reduced by averaging the hundreds of frames acquired during a single I_0 scan. However, in the present case the noise in the filtered region can be substantial due to the reduced intensity. Furthermore, with dynamic filtration I_0 varies with angle so there are not multiple realizations available for averaging. Thus, we performed non-linear denoising on the I_0 data prior to the log normalization. Anisotropic diffusion was used as it typically provides better edge preservation than traditional smoothing filters.

Alignment of the I (object) and I_0 (air) scans began by selecting the closest angular match. From scan to scan variability analysis we expect this to align the filter edges to within a few pixels. To correct residual misalignment, the I_0 frame was split in the middle of the aperture

and each half independently registered to the I frame as follows. The normalized projection I_0/I was integrated along $\vec{u}$ (the direction parallel to the blade edges, as shown in Fig. 4) to enhance the blade edges and reduce anatomical structure. The summed absolute derivative of this integral over a region containing the blade edges was then minimized by splitting the I_0 image and slightly shifting the two halves in the $\vec{u}$ direction, effectively moving the blade edges. This cost function demonstrates a single, well defined minimum, even for images of a complex structure like the RANDO head phantom. As indicated by the flow chart the cost was evaluated after additional corrections, on the best estimate of sinogram.

Beam Quality Correction—If we include the energy dependence of the x-ray attenuation, the imaging model of can be more accurately written as

$$I = I_0 \int S(E) e^{-\int \mu(\vec{r}, E) dl} dE \quad (4)$$

where $S(E)$ is the normalized spectrum and μ is now a function of energy as well as position. The linear form of Equation (1) comes from assuming the measured attenuation coefficient represents an average energy ($\bar{E}$). This approximation generally performs well for typical CT imaging, though it can introduce beam hardening artifacts which can present as cupping or dark shading between high density region arising from the hardening in the spectrum as it passes through the object. In our case, the copper filters used for the intensity weighting cause substantial hardening of the beam spectra as shown in Fig. 7. In the linear model this results in two different average energies where $\bar{E}_{\text{filtered}} \gg \bar{E}_{\text{open}}$. If left uncorrected this leads to a bias in the images with lower values in the filtered regions of the image. Additionally, the consistent, sharp transition leads to truncation like artifacts corrupting the inner region. However, we find that the following approximation

$$\mu_A(\bar{E}_{\text{open}}) \approx \alpha \mu_A(\bar{E}_{\text{filtered}}) \quad \text{where: } \alpha = \frac{\mu_{\text{water}}(\bar{E}_{\text{open}})}{\mu_{\text{water}}(\bar{E}_{\text{filtered}})} \quad (5)$$

where the subscript on μ indicates a specific material, is valid for most materials of interest, particularly water and soft tissue. However there is some error in bone resulting in a bias in the image values for bone in the filtered regions of the image.

From Equation (5) it follows that the measured intensity weighted sinogram can be corrected for the harder beam under the filters to approximate a sinogram acquired with a uniform, unfiltered beam simply by scaling the data in the filtered regions as shown below.

$$p_{bq}(u, v; \theta) = w_{bq}(u, v; \theta) \times p(u, v; \theta) \\ \text{where: } w_{bq}(u, v; \theta) = \begin{cases} 1 & \text{if } u, v; \theta \in \{\text{open}\} \\ \alpha & \text{if } u, v; \theta \in \{\text{filtered}\} \end{cases} \quad (6)$$

The value of the beam quality (α) correction parameter can be estimated from knowledge of the filtered and unfiltered beam spectra and the NIST x-ray attenuation tables [23]. We further refined this estimate by scanning a homogeneous solid plastic cylinder with an off center, elliptical ROI for which the ideal projections could be modeled analytically. The parameter was then optimized to minimize the difference between the IWROI measured projection data and the modeled data.

Transition region smoothing—In cases where there was residual error resulting in unwanted structure in the transition between the filtered and unfiltered region an optional smoothing operation was applied. The region in the sinogram pixels from the identified filter edge position was replaced with a spline fit spanning the gap.

3.7. Monte carlo dose modeling

To obtain quantitative estimates of dose and scatter reduction with conformal and IWROI imaging, we performed simulations using the EGSnrc MonteCarlo system [17, 24]. A model of the OBI source was created with the BEAMnrc user code [25], using geometry from Varian [26]. The BEAMnrc component modules XTUBE, SLABS and JAWS were used to model the x-ray tube's W95-Rh5 rotating anode, glass/oil/polycarbonate composite exit window, Al prefilter, upper and lower lead OBI blades, Cu or Pb dynamic collimator blades, and intervening air gaps. The dynamic blades were located 25 cm from the anode. Different filter blade materials and IWROI aperture sizes could then be modeled by changing the material and geometry parameters in the relevant component module. The origin of the BEAMnrc coordinate system was located 1 cm above the center of the anode with the Z-axis along the output beam direction. A 0.5 mm diameter electron beam at energy 125 keV impinged on the center of the anode at an angle of 12° above the XY plane and in the XZ plane. Selected properties of the simulations are given in Table 1.

Three-dimensional dose distributions were computed using the DOSXYZnrc user code. For dynamic ROI imaging the incident beam size and position vary with gantry angle, so a single phase space source could not be used to simulate this technique. Instead we used a full BEAMnrc simulation for each projection angle, each with its own asymmetric jaw settings conforming to the ROI. It was found that simulating 1 million incident photons at each of 180 equally spaced angles gave satisfactory results, with 1 to 2% dose uncertainty within the ROI and 2 to 3% elsewhere. Conveniently DOSXYZnrc scores dose in Gray per incident particle in the original simulation, i.e. per electron incident on the anode. Thus the results were readily scaled to dose per milliamp-second and thence to integrated mAs for the CBCT scans:

$$D_{\text{CBCT}} (\text{mGy}) = D_{\text{CBCT}} (\text{Gy/electron}) \div 1.602 \cdot 10^{-19} (\text{A} \cdot \text{sec/electron}) \times I (\text{mA}) \times t (\text{sec/projection}) \times N (\text{projections/scan}) \quad (7)$$

3.7.1. MC model dose validation—Monte Carlo doses were compared to film measurements in a transverse plane of the RANDO phantom for several irradiation conditions. The measurement utilized Gafchromic XR-QA2 film which was calibrated using

a 125 kVp beam from a small animal irradiator (X-Rad225Cx, Precision X-ray, Inc) at 12 known doses from 0 to 200 mGy. The calibration and RANDO experimental films were scanned (5 scans per film, averaged) on an Epson Perfection 10000XL color scanner and saved in TIFF files with 16 bits per color channel. Scanned images were converted to dose using a multichannel method based on the ratio of signals in the red and green channels (with XR-QA2 film the blue channel has no dose response). From the $2.5 \times 2.5 \times 2.5$ mm DOSXYZnrc grid, a transverse plane was chosen reasonably close to the film plane, although small errors in both in plane and out of plane rotations due to setup differences remained. The agreement between film and Monte Carlo doses in the interior was excellent, while near the surface the film doses appeared a few mGy lower, which may be due to the difficulty of completely eliminating air gaps between the film and the RANDO slabs. [Images omitted in the interest of space.]

4. Results

4.1. Conformal ROI images

Conformal imaging ROIs were chosen as peripheral regions so as to be reconstructable with chord-based BPF and also to be deliverable with the collimator hardware. Reconstruction results are shown in Fig. 8. The region of the ear in the RANDO scan and the small section of thigh on the left edge of the chicken were not intended to extend outside the ROI. This indicates that insufficient margins were used to account for our setup uncertainties.

Generally, the ROIs from the conformal data sets are well recovered; however there are two artifacts evident in the images. The first is an intensity drop out in the image, seen as a dark streak along the edge in RANDO and the upper right corner of the chicken image. The other is the bright section on the outer edge. Both are a result of misalignment of the collimator edges and could be avoided by expanding the collimator trajectory out from the reconstruction ROI.

While this technique was able to accurately recover the ROI it has several major limitations. The ROI must be able to be filled with untruncated chords, line segments connecting points on the circular trajectory, effectively limiting it to peripheral ROIs such as those shown here. This is not practical for IGRT where many of the targets are situated deep within the body. Also, the complete lack of image outside the ROI means that the ROI must include all structures of interest for both targeting and overall alignment, likely making it much larger and reducing the dose sparing potential. Finally, both the requirement for untruncated chords and the lack of image recovery outside the ROI make the technique rather unforgiving to setup errors or other changes in the patient anatomy.

4.2. diWROI images

For direct comparison to the conformal ROI technique, the same peripheral ROIs were scanned with the copper filters rather than radio-opaque blades. The RANDO head phantom result is shown in Fig. 9 along with the conventional CBCT result. Qualitatively the ROI is just as well recovered in the diWROI image as the conformal image in Fig. 8 with the additional benefit of a useful image of the rest of the head. In the outer region the noise is

higher and there are some pronounced streak artifacts, but it contains sufficient information for patient alignment.

The chicken phantom with enlargements showing the soft tissue structures is shown in Fig. 10. The ROI is again well recovered. In the ROI enlargement there are ring artifacts in the dIWROI that are not present in the conventional image. This is most likely due to shifting the I_0 frame to match the blade edges for the log-normalization which means that the detector structure (due to non uniform pixel response, etc) no longer divides out. This could be corrected with a separate detector response correction applied prior to the log-normalization step.

The results for the interior elliptical ROI are shown in Fig. 11 demonstrating again a high quality image within the ROI and lower quality but useful image outside. Note that this interior ROI could not have been reconstructed using conformal ROI imaging.

4.3. Monte carlo dose results

The Monte Carlo calculated dose distributions for the RANDO phantom imaging studies were normalized by the dose from an unfiltered scan and these distributions are shown in Fig. 12 along with histograms. For both ROIs most of the imaging dose is deposited in the target and the dose to the rest of the head is reduced. From the histogram for the peripheral ROI we see that the dose in the rest of the head is reduced to 55–80% of the open field dose. For the interior elliptical ROI the sparing is even greater, with most of the outer region in the range of 20–60% of the open field dose. This difference is due to the size and position of the ROI, with the larger peripheral ROI causing more unfiltered beam exposure of non-ROI tissue. Note that since the x-ray transmission through the Cu filter blades is less than 4%, even for peripheral ROIs where true conformal ROI imaging is possible the IWROI technique still accomplishes almost the same dose reduction to non-ROI regions [data not shown]. Imaging dose within the ROI is also lower in the dIWROI images, due to reduced scatter from the smaller volume exposed to the full x-ray beam intensity. This effect is greater (i.e., lower dose within the ROI) as the ROI size decreases, since more volume is spared from the full-intensity beam thereby reducing the source for scatter dose to the ROI. This reduction in irradiated volume will also lead to reduced scatter contamination in the projection images, and in the reconstructed image volume.

5. Conclusions

Region of interest imaging has the potential to allow dose reduction by confining the imaging beam only to the region within which an image is needed, thus sparing imaging dose to tissues presumably away from the target in image-guided radiotherapy. However, nonperipheral regions are subject to the interior problem and are not amenable to correct reconstruction from a true conformal set of projections, even with advanced reconstruction techniques such as the chord-based BPF method.

The work presented here demonstrates the feasibility of dynamic intensity-weighted ROI imaging in conebeam CT to reduce imaging dose to the patient for image guided therapies while still allowing proper reconstruction. Spatially varying illumination can be

accomplished with a partially transmitting aperture that reduces incident x-ray intensity outside the ROI, while fully irradiating the ROI at each view. The design and construction of an electronically controlled, motor driven collimator device with partially transmitting copper blades to accomplish dynamic intensity-weighted ROI illumination was described. The device was shown to be sufficiently accurate for practical use in ROI tracking illumination strategies as used here. Appropriately designed data preprocessing software and simple filtered back-projection reconstruction (FDK) were used to reconstruct images with high quality within the ROI, and reduced quality elsewhere. Such an image is typically sufficient for patient setup and soft tissue target localization. Dose reduction was quantitatively assessed using Monte Carlo simulations, showing significant decrease outside the ROI compared to full-fan irradiation and reduced scatter contamination.

Intensity-weighted region-of-interest imaging has the advantage of allowing use of standard filtered-backprojection reconstruction. Appropriate data preprocessing steps to account for the spatially varying incident intensity, beam hardening by intensity-weighting filters and possibly edge matching to account for variability in blade positions between I and I_0 scans need to be applied but these are straightforward processes. The major difficulty is the need for a suitable dynamic intensity-weighting collimator such as the dynamic copper blade mechanism described here. Commercial systems such as the Varian On-Board Imager do not support dynamic field shaping during a scan, and in any event would need to be designed with a second set of partially attenuating blades. One possibility if the existing opaque blades were capable of motion during a scan would be to modulate the intensity by sliding the blades back and forth in a suitable pattern so the ROI is always illuminated with full fluence and the rest of the volume with only partial fluence. This strategy would be unlikely to yield a dataset reconstructible with simple FBP, however; advanced iterative algorithms capable of handling few-view datasets would likely be required. Yet another possibility with dynamic but opaque blades would be to perform two full scans, the first with very low fluence and full field illumination, the second with high fluence and conformal illumination. Merging these two datasets would produce a set of projections equivalent to those we have acquired with the dynamic intensity-weighting collimator, which could be reconstructed following the procedure we have outlined in this paper. This is essentially the method described in [27]. Merging the two projection sets would require view-by-view image registration to match the projected anatomy and could be compromised by respiratory or other patient motion between the two scans, whereas in the technique reported in the present paper the high and low fluence portions of each projection are acquired simultaneously. An advantage of the two-scan approach would be that since there is no filtering of the low-intensity beam, no adjustment for spectral differences would be needed as in the present work. A disadvantage would be scanning time, since two complete scans would be taken. Time is of the essence in radiotherapy delivery, so adding another minute to image acquisition could be a significant impediment to adoption of the method.

Parsons and Robar [28, 29] have developed an innovative iris-type dynamic collimator for ROI imaging in IGRT. At present they have utilized it for true conformal imaging, in which they only attempt to recover an accurate image of the specified volume-of-interest (VOI). As such, it does not provide as much information of the surrounding anatomy as the dIWIROI method presented here. The blades of the iris collimator described in [28] are copper with a

thickness of 6.4 mm, twice the thickness we used; thus they will have a nonzero, though small (roughly 0.2%) transmission. In principle there is no reason the iris collimator cannot be used for IWROI imaging, although its relatively complicated blade structure would make edge matching as described above for the I_0 correction quite challenging. The iris blades overlap variably as the aperture changes, and these sharp transitions would also need to be precisely matched between I and I_0 images.

Graham [30, 31] and Bartolac [32, 33] have reported on “Intensity-Modulated” cone-beam CT which is a similar concept to the IWROI methods presented here. Their approach has been a theoretical study of the relationship of incident fluence and image SNR and to generate continuous fluence modulation patterns to achieve a specified SNR distribution. This approach provides a continuous trade off between the local imaging dose and image quality; however the fluence patterns they require would be difficult to generate. Combining their theoretical planning methods with the dynamic collimator and real data reconstructions presented here could produce a general framework for patient specific, task prescribed, variable image quality with reduced and targeted imaging dose.

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References

1. Nuclear Regulatory Commission, Standards for protection against radiation. Washington, DC: US Government Printing Office; 1991.
2. Murphy MJ, Balter J, Balter S, BenComo JA Jr, Das IJ, Jiang SB, Ma CM, Olivera GH, Rodebaugh RF, et al. The management of imaging dose during image-guided radiotherapy: Report of the AAPM Task Group 75. *Med Phys*. 2007; 34:4041–4063. [PubMed: 17985650]
3. Greess H, Wolf H, Baum U, Lell M, Pirkel M, Kalender WA, Bautz WA. Dose reduction in computed tomography by attenuation-based on-line modulation of tube current: Evaluation of six anatomical regions. *Eur Radiol*. 2000; 10:391–394. [PubMed: 10663775]
4. Kalender, WA., Wolf, H., Suess, C., Gies, M., Greess, H., Bautz, WA. *Eur Radiol*. Vol. 9. Springer-Verlag; 1999. Dose reduction in CT by on-line tube current control: Principles and validation on phantoms and cadavers; p. 323-328.
5. La Rivière PJ. Penalized-likelihood sinogram smoothing for low-dose CT. *Med Phys*. 2005; 32:1676–1683.
6. Wang J, Lu H, Wen J, Liang Z. Multiscale penalized weighted least-squares sinogram restoration for low-dose x-ray computed tomography. *IEEE Trans Biomed Eng*. 2008; 55:1022–1031. [PubMed: 18334394]
7. Wang J, Li TF, Lu H, Liang Z. Noise Reduction for Low-Dose Single-Slice Helical CT Sinograms. *IEEE Trans Nucl Sci*. 2006; 53:1230–1237. [PubMed: 16932806]
8. Beekma FJ, Kamphuis C. Ordered subset reconstruction for x-ray CT. *Phys Med Biol*. 2001; 46:1835–1844. [PubMed: 11474928]
9. Manglos SH, Gagne GM, Krol A, Thomas FD, Narayanaswamy R. Transmission maximum-likelihood reconstruction with ordered subsets for cone beam CT. *Phys Med Biol*. 1995; 40:1225–1241. [PubMed: 7568379]
10. Nuyts J, De Man B, Dupont P, Defrise M, Suetens P, Mortelmans L. Iterative reconstruction for helical CT: A simulation study. *Phys Med Biol*. 1998; 43:729–737. [PubMed: 9572499]

11. LaRoque SJ, Sidky EY, Pan X. Accurate image reconstruction from few-view and limited-angle data in diffraction tomography. *J Opt Soc Am A*. 2008; 25:1772–1782.
12. Sidky EY, Kao CM, Pan X. Accurate image reconstruction from few-views and limited-angle data in divergent-beam CT. *J X-ray Sci Tech*. 2006; 14:119–139.
13. Sidky EY, Kao CM, Pan X. Effect of the data constraint on few-view, fan-beam CT image reconstruction by TV minimization. *IEEE Nuclear Science Symposium Conference* (. 2006:2296–2298.
14. Feldkamp LA, Davis LC, Kress JW. Practical cone-beam algorithm. *J Opt Soc Am A*. 1984; 1:612.
15. Cho, S., Pearson, E., Pelizzari, CA. Noise analysis in intensity-weighted region-of-interest imaging for cone-beam CT. In: Samei, E., Hsieh, J., editors. *SPIE Medical Imaging...* 7258. SPIE; 2009. p. 725807–725807–9
16. Cho S, Bian J, Pelizzari CA, Chen CT, He TC, Pan X. Region-of-interest image reconstruction in circular cone-beam microCT. *Med Phys*. 2007; 34:4923–4933. [PubMed: 18196817]
17. Pearson, E., Pan, X., Pelizzari, CA. *Med Phys*. Vol. 41. American Association of Physicists in Medicine; 2014. TH-A-18C-10: Dynamic intensity weighted region of interest imaging; p. 542-542.
18. Pearson, E. PhD Thesis. The University of Chicago; 2013. Development and application of advanced cone-beam CT acquisition strategies for image-guided therapies.
19. Pearson, E., Al-Hallaq, HA., Pan, X. *IEEE Nuclear Science Symposium and Medical Imaging Conference (NSS/MIC)*. IEEE; 2011. Dynamic region-of-interest cone-beam CT for image-guided postmastectomy radiotherapy; p. 3187-3190.
20. Pan, X., Xia, D., Zou, Y., Yu, L. *Phys Med Biol*. Vol. 49. IOP Publishing; 2004. A unified analysis of FBP-based algorithms in helical cone-beam and circular cone-and fan-beam scans; p. 4349-4369.
21. Pan X, Zou Y, Xia D. Image reconstruction in peripheral and central regions-of-interest and data redundancy. *Med Phys*. 2005; 32:673–684. [PubMed: 15839339]
22. Zou, Y., Pan, X., Sidky, EY. *Phys Med Biol*. Vol. 50. IOP Publishing; 2005. Image reconstruction in regions-of-interest from truncated projections in a reduced fan-beam scan; p. 13-27.
23. Hubbell, JH., Seltzer, SM. *Tables of X-ray mass attenuation coefficients and mass energy-absorption coefficients (version 1.4)*. National Institute of Standards and Technology; 2004.
24. Sadinski, M., Pearson, E., Pelizzari, CA. Dose Reduction Intensity Weighted Region of Interest and True Region of Interest Conebeam CT Imaging: A Monte Carlo and Phantom Study; *RSNA 98th Annual Meeting*; Chicago, IL. 2012.
25. Rogers, DWO. *Med Phys*. Vol. 22. American Association of Physicists in Medicine; 1995. BEAM: A Monte Carlo code to simulate radiotherapy treatment units; p. 503-524.
26. On-Board Imager and PortalVision Monte Carlo Data Package: Dwg. No. 100040466-03. Varian Medical Systems. 2008
27. Chityala, RN., Hoffmann, KR., Rudin, S., Bednarek, DR. *SPIE Medical Imaging*. Vol. 5745. SPIE; 2005. Region of interest (ROI) computed tomography (CT): Comparison with full field of view (FFOV) and truncated CT for a human head phantom; p. 583-590.
28. Parsons, D., Robar, JL. *Med Phys*. Vol. 42. American Association of Physicists in Medicine; 2015. An investigation of kV CBCT image quality and dose reduction for volume-of-interest imaging using dynamic collimation; p. 5258-5269.
29. Parsons, D., Robar, J. *Med Phys*. Vol. 41. American Association of Physicists in Medicine; 2014. TH-A-18C-11: An investigation of KV CBCT image quality and dose reduction for volume-of-interest imaging using dynamic collimation; p. 542-542.
30. Graham SA, Siewerdsen JH, Keller H, Moseley DJ. TU-D-I-611-04: Intensity-modulated cone-Beam CT for patient-specific distribution of SNR. *Med Phys*. 2005; 32:2092.
31. Graham, SA., Siewerdsen, JH., Jaffray, DA. Intensity-modulated fluence patterns for task-specific imaging in cone-beam CT. In: Hsieh, J., Flynn, MJ., editors. *Medical Imaging*. Vol. 6510. International Society for Optics and Photonics; 2007. p. 651003–651003–9
32. Bartolac S, Graham SA, Siewerdsen JH, Jaffray DA. TU-A-201B-01: An Inverse Planning Method for Intensity Modulated Computed Tomography. *Med Phys*. 2010; 37(6):3372.

33. Bartolac S, Graham SA, Siewerdsen JH, Jaffray DA. Fluence field optimization for noise and dose objectives in CT. *Med Phys*. 2011; 38:S2–S17.

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Fig. 1. Intensity weighted ROI imaging concept. ROI is illuminated with full intensity beam (red), remainder of anatomy with intensity attenuated by filters (copper collimator blades, in our implementation).

Fig. 2.
(Left) Collimator mounted on the kV imager housing. (Right) Control electronics attached to the face of the gantry.

Fig. 3. Measured (blue) and model predicted (green dashed) positions for an arbitrary waveform input.

Fig. 4.

The left panel shows a sample radiograph with one blade extended roughly halfway. The upper right panel shows the histogram of the image with the two peak values automatically identified. The edge threshold (orange line) is set as the midpoint of these two peaks and the blade position is determined to be where the middle row profile crosses this value, lower right panel.

Fig. 5.
The head section of the RANDO Man phantom (left), and the Chicken phantom (right) setup for planning scan on the Philips Brilliance Big Bore CT.

Fig. 6.
Flow chart depicting the data correction chain used for the diWROI imaging studies.

Fig. 7. Normalized beam spectra for the unfiltered and filtered regions of the x-ray beam, calculated with Monte Carlo.

Fig. 8. BPF reconstructions of the RANDO (top) and chicken (bottom) phantoms. Left column is the “reference” reconstruction from full illumination with fan chords. Right is the ROI reconstruction from conformal illumination with the non-overlapping chord arrangement. Display windows are [0.1, 0.3] for RANDO and [0.15, 0.25] for the chicken.

Fig. 9.
(a) The reference CBCT image of the RANDO head phantom. (b) the dIWROI image with the right-anterior peripheral ROI of Fig. 8.

Fig. 10.

Reconstructions of chicken phantom, peripheral ROI. Top, reference reconstruction from full data; bottom, diWROI reconstruction. The right column is an enlargement of a 24.4 mm×24.4 mm region within the ROI. The enlargement region was chosen to show low contrast muscle boundaries in the thigh. Display window [0.17, 0.23] left and [0.18, 0.24] right.

Fig. 11.

Left: reference, unfiltered image of the RANDO phantom head. Center: reconstruction from uncorrected dIWROI data for elliptical interior ROI. Right: dIWROI reconstruction with data corrections applied. Insets show enlargements of fine bone structure in the ROI. Slight difference in orientation between reference and dIWROI slices is due to imperfect phantom repositioning between scans.

Fig. 12.

Monte Carlo calculated dose for the diWROI scans as a ratio to the unfiltered reference scan dose. Top row is for the peripheral ROI and bottom row the interior elliptical ROI. Right column shows dose distribution on central plane and left column the volume histogram.

Table 1

Monte Carlo simulation parameters

Parameter	Value(s)
Incident electron energy	125 keV
Beam geometry	0.5 mm circular parallel beam with -12° angle of incidence
Global electron cutoff ECUT	512 keV
Global photon cutoff PCUT	1 keV
Physical processes and variance reduction techniques	<ul style="list-style-type: none">• Directional bremsstrahlung splitting• Bremsstrahlung cross-section enhancement (anode)• Electron impact ionization• Atomic relaxation• Bound Compton scattering• Rayleigh scattering