Tomographic image via background subtraction using an x-ray projection image and a *priori* computed tomography

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Kilovoltage x-ray projection images (kV images for brevity) are increasingly available in image guided radiotherapy (IGRT) for patient positioning. These images are two-dimensional (2D) projections of a three-dimensional (3D) object along the x-ray beam direction. Projecting a 3D object onto a plane may lead to ambiguities in the identification of anatomical structures and to poor contrast in kV images. Therefore, the use of kV images in IGRT is mainly limited to bony landmark alignments. This work proposes a novel subtraction technique that isolates a slice of interest (SOI) from a kV image with the assistance of a *priori* information from a previous CT scan. The method separates structural information within a preselected SOI by suppressing contributions to the unprocessed projection from out-of-SOI-plane structures. Up to a five-fold increase in the contrast-to-noise ratios (CNRs) was observed in selected regions of the isolated SOI, when compared to the original unprocessed kV image. The tomographic image via background subtraction (TIBS) technique aims to provide a quick snapshot of the slice of interest with greatly enhanced image contrast over conventional kV x-ray projections for fast and accurate image guidance of radiation therapy. With further refinements, TIBS could, in principle, provide real-time tumor localization using gantry-mounted x-ray imaging systems without the need for implanted markers. © 2009 American Association of Physicists in Medicine. [DOI: 10.1118/1.3193525]

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I. INTRODUCTION

With the prevalence of highly conformal radiation treatment techniques, such as intensity-modulated radiotherapy (IMRT),^{1–4} stereotactic radiosurgery/radiotherapy,⁵ and intensity modulated arc therapy,⁶ radiotherapy increasingly demands high accuracy on patient positioning and target localization prior to therapeutic beam delivery.

The recently developed onboard imaging (OBI) system with kilovoltage cone beam CT (CBCT) capability allows two-dimensional (2D) and three-dimensional (3D) image guidance for treatment positioning.^{7,8} CBCT images exhibit sufficient soft-tissue contrast for online 3D soft-tissue localization. However, the excessive scanning time and imaging dose may discourage daily uses of CBCT in IGRT—such as patient positioning.^{9,10} Kilovoltage (kV) x-ray radiographic imaging is more often employed in daily imaging guidance because of its low dose and quick snapshot nature. However, projecting a 3D object onto an imaging plane disposes of the depth information and reduces the possibility of distinguishing anatomic structures in the 2D image. Because superimposed information is displayed on a 2D projection, the anatomical structures contained within a slice of interest (SOI) appear more or less shadowed by the thickness of the tissue above and below. This feature of 2D kV projection images limits their use for soft-tissue localization. Recently, there has been research interest on patient positioning using alternative x-ray imaging modality aiming to achieve lower im-

aging dose and faster acquisition such as digital tomosynthesis, which provides decent soft-tissue contrast by scanning in about 30° angular range.^{11,12}

In this work, we propose a novel technique called tomographic image via background subtraction (TIBS). TIBS operates on a conventional kV image by removing background information—i.e., not belonging to the SOI—from the x-ray path with the assistance of a previously acquired CT image. TIBS is able to isolate anatomical structures included within any preselected slice, greatly decreasing ambiguities and increasing contrast of the 2D unprocessed kV images.

II. METHODS AND MATERIALS

II.A. Principle of the TIBS technique

TIBS is a digital subtraction technique in concept similar to digital subtraction angiography (DSA)¹³ and computed tomography angiography (CTA).¹⁴ DSA and CTA images are produced by collecting data before and after the administration of a contrast agent and subtracting the results, while TIBS aims to isolate an SOI by subtracting a *priori* background information from kV images.

Consider an x-ray point source and a flat-panel x-ray detector. The detected intensity $I(\vec{s})$ from primary plus scattered radiation at detector pixel location $\vec{s}$ is given by

$$I(\vec{s}) = I_0(\vec{s})e^{-P(\vec{s})}(1 + \text{SPR}), \quad (1)$$

where $I_0(\vec{s})$ is the incident intensity in the absence of scan object, $e^{-P(\vec{s})}$ is the attenuation of primary beam only, and

FIG. 1. Schematic illustration of TIBS technique. TIBS is obtained by subtracting the DRR of the background from the kV image.

SPR is the scatter-to-primary ratio.¹⁵ The primary beam attenuation is equal to the line integral of the linear attenuation coefficient $\mu(\vec{r})$,

$$P(\vec{s}) = \int_{\vec{r} \in L} \mu(\vec{r}) d\vec{r}, \quad (2)$$

where $\vec{r}$ is the voxel location vector of a point within the scan object and L is the x-ray path connecting the x-ray source, the points in the object traversed by the x-ray beam and the detector pixel position (a straight line connecting $\vec{r}$ and $\vec{s}$). The kV images $K(\vec{s})$, as well as projections for CBCT reconstruction in our paper, are calculated as the negative logarithm of the transmission,

$$K(\vec{s}) = -\ln \frac{I(\vec{s})}{I_0(\vec{s})}. \quad (3)$$

In the ideal case where $\text{SPR}=0$, $K(\vec{s})$ is equal to $P(\vec{s})$. Otherwise the relationship between these two quantities requires the precise knowledge on SPR, which may not be available. If uncorrected, x-ray scatter will cause cup and streak artifacts and inaccurate CT number in reconstructed CT images.^{14,15} High scattered radiation levels are especially problematic in flat-panel based CBCT because beam collimation is extremely challenging for large cone angles. In the Varian OBI system we used in our study, a significant portion of scatter is rejected by the use of a 10:1 antiscatter grid and a 50 cm air gap between scan object and detector.

Under the assumption that the position of the patient anatomy at the time of kV imaging is not significantly distorted so that it can be aligned with that of the previous CBCT scan, the background contents can be considered invariant between the time of CBCT scan and kV imaging. In the above situation, the TIBS technique consists of the following steps, as illustrated in Fig. 1.

- (1) Volumetric CBCT images $\mu_{\text{eff}}(\vec{r})$ are reconstructed from CBCT projection data using our in-house FDK algorithm.¹⁶ Here $\mu_{\text{eff}}(\vec{r})$ represents the “effective” x-ray linear attenuation coefficient map that contains cup and streak artifacts and other inaccuracies resulting from

scatter and beam hardening. A SOI is isolated from the reconstructed 3D CBCT volume, and the voxels not belonging to the SOI are considered as “background” (BKG). A 3D BKG dataset is created by setting those $\mu_{\text{eff}}(\vec{r}) \in \text{SOI}$ equal to zero.

- (2) Numerically computed projections (also known as digital reconstructed radiographies—DRRs) of SOI and BKG are generated according to Eqs. (4) and (5), respectively,

$$\text{DRR}_{\text{SOI}}(\vec{s}) = \int_{\vec{r} \in L \cap \text{SOI}} \mu_{\text{eff}}(\vec{r}) d\vec{r}, \quad (4)$$

$$\text{DRR}_{\text{BKG}}(\vec{s}) = \int_{\vec{r} \in L \cap \text{BKG}} \mu_{\text{eff}}(\vec{r}) d\vec{r}, \quad (5)$$

where $\text{DRR}_{\text{BKG}}(\vec{s})$ is to be used to generate TIBS and $\text{DRR}_{\text{SOI}}(\vec{s})$ is used as a reference image for verification purposes.

- (3) Subtract $\text{DRR}_{\text{BKG}}(\vec{s})$ from kV image to obtain the TIBS image of SOI as shown in Eq. (6), where $\text{TIBS}_{\text{SOI}}(\vec{s})$ denotes background-subtracted image. Under ideal situations, $\text{TIBS}_{\text{SOI}}(\vec{s})$ should equal to $\text{DRR}_{\text{SOI}}(\vec{s})$.

$$\text{TIBS}_{\text{SOI}}(\vec{s}) = K(\vec{s}) - \text{DRR}_{\text{BKG}}(\vec{s}) \quad (6)$$

In practice, the kV imaging setup may not be perfectly inline with that of CBCT scan. In such cases, an image registration step needs to be inserted between steps (2) and (3) to align DRR_{BKG} with kV image before the subtraction of the two.

II.B. Phantom experiment evaluation

TIBS was implemented using the OBI system on a linear accelerator (Trilogy, Varian Medical Systems, Palo Alto, CA). Two phantoms (phantoms 1 and 2) were designed with six polymethyl methacrylate (PMMA) rods and two disk modules of a disassembled Catphan® phantom (The Phantom Laboratory, Salem, NY). The six PMMA rods are all 15 cm long and 1.2 cm in diameter; and the two disk modules are both 15 cm in diameter and 2.4 and 4 cm thick, respectively. For phantom 1, a sensitometry module of the Catphan® phantom and the six rods were placed in a water filled plastic bucket 21 cm in diameter and 14 cm in height. As shown in Fig. 2, three rods were placed on top of the disk module along the couch [superior-inferior (SI)] and the other three were placed below the disk perpendicular to the couch [left-right (LR)]. The imaging isocenter was placed around the center of the disk module. The sensitometry module contains eight cylindrical inserts made of different materials. The projections of the eight cylinders in the anterior-posterior (AP) direction were circular regions on the kV image and on the subtracted slice image. These regions were used to measure relative contrast-to-noise ratios (CNR). CNR is a simple and objective measure of the detectability of certain structures with uniform intensity. It is defined by the formula: $\text{CNR} = |S_{\text{in}} - S_{\text{out}}| / \sqrt{\sigma_{\text{in}}^2 + \sigma_{\text{out}}^2}$, where S_{in} and

FIG. 2. Design of phantom. Six PMMA rods and a disk module of the Catphan® phantom (The Phantom Laboratory, Salem, NY) are placed in a water bucket.

S_{out} are the averaged image pixel intensities within a given region of interest (ROI) and a uniform background region outside the ROI, respectively; σ_{in} and σ_{out} are the standard deviations of noise inside and outside the ROI, respectively.¹⁷ Phantom 2 was designed in the same way as phantom 1 except that a low-contrast disk module replaced the sensitometry one to test the low-contrast visibility in TIBS images.

Projection data for CBCT reconstruction were acquired using a $40 \times 30 \text{ cm}^2$ Paxscan® 4030CB onboard flat-panel imager (OBI, Varian Medical Systems, Palo Alto, CA). Each set of projection data contains 1024×768 elements where each element covers a $0.388 \times 0.388 \text{ mm}^2$ area. A CBCT acquisition consisted of ~ 630 evenly distributed view angles over a complete circle of 360° , with a “standard” technique of 125 kVp, 80 mA, and 15 ms, a source-to-axis distance (SAD) of 100 cm, and source-to-detector distance (SDD) of 150 cm. All CBCT scans were operated under the full-fan (i.e., 25 cm diameter field of view or less) CBCT mode with a 10:1 antiscatter grid and a bow-tie filter mounted. The kV images we used for our study are selected among the projections acquired during the CBCT scan. By acquiring the kV images with the same technique, filtering condition, and geometric setup, we eliminate any misalignments between CBCT and kV images as well as possible differences in the data normalization. Figure 3(a) shows the kV image at 0° x-ray source angle (AP direction). In order to demonstrate the effects of geometrical misalignment on TIBS results, we also generated intentional-shifted kV image set before applying the TIBS technique.

The projection data from the CBCT scans were subjected to dark/flood-field calibration and bad-pixel correction. These projections were then divided by the projection of a normalization cylindrical phantom (NORM-phantom) whose density $\mu_0(\vec{r})$ is expected to be close to that of the scan object. This normalization procedure is essentially equivalent to Eq. (3) where $I_0(\vec{s})$ is the NORM-phantom attenuated intensity instead of “open-field” reading, and it helps reduce the effects of scatter and beam hardening.¹⁸ Finally the natural logarithm was performed on the ratio of the intensities to form the projection data $K(\vec{s})$. CBCT voxels corresponding to the relative linear absorption coefficients ($\mu_{\text{eff}} - \mu_0$) were

FIG. 3. Phantom 1: kV image (a) and TIBS of three coronal slices at different horizontal positions with thicknesses of 1.3, 1.3, and 2.4 cm for (b), (c), and (d), respectively.

reconstructed using our in-house FDK algorithm with Blackman filter.¹⁶ (Unless otherwise specified, kV images and CBCT projections are after NORM-phantom normalization and logarithm transformation. Reconstructed CBCT voxel values represent relative absorption coefficients). Our CBCT volume matrix contains 400 axial slices with 0.55 mm slice thickness; each slice is a matrix of 600×600 pixels with size of $0.46 \times 0.46 \text{ mm}^2$. Neither x-ray scattering nor beam hardening was corrected in our reconstruction algorithm. An SOI was selected in the reconstructed CBCT volume, and the BKG was created by setting the SOI voxel values to zero. Voxels belonging to BKG were projected onto a 1024×768 simulated flat-panel detector using a standard ray-driven forward projection method to generate DRR_{BKG} . The subtracted result TIBS_{SOI} was obtained by subtracting DRR_{BKG} from the kV image. Theoretically, the resulting TIBS image contains only the information in SOI and closely mimics the reference image DRR_{SOI} .

II.C. Head and neck study

TIBS was tested with a head and neck case. The CBCT scan was operated under the full-fan mode with bow-tie filter mounted. CBCT projection data were acquired at ~ 600 different view angles using the Paxscan® 4030CB imager.

First the feasibility of TIBS was demonstrated under ideal condition, where setup errors and data normalization mismatches were absent between CBCT and kV images. Two projections were retrospectively selected from the same CBCT dataset to be used as our AP and LR direction kV radiographic images. In this ideal and somewhat unrealistic scenario, the TIBS technique was repeated on five sets of CBCT data that correspond to weeks 1–5 of the treatment course. All five CBCT datasets were normalized to the NORM-phantom and logarithm-transformed, and kV images of each week were taken from their corresponding same week CBCT dataset.

We also tested the feasibility of TIBS in a more realistic scenario, where week 2 CBCT was used as *a priori* information and projections of week 3 CBCT were used as the LR kV images acquired on the treatment day (1 week after previous CBCT was taken). Geometrical mismatch existed between the kV images and CBCT dataset due to setup errors. A 2D rigid-body registration between the week 3 kV image and week 2 DRR image was applied to correct setup errors before applying subtraction. Image registration software FSL¹⁹ was used and the 2D-2D three-parameter rigid registration method was selected in the FLIRT function,²⁰ which considers two translation components and one rotation angle within the 2D imaging plane.

III. RESULTS

III.A. Phantom study

Figure 3(a) shows the kV image of phantom 1, in which the six rods and the sensitometry module are overlapped in the image. Three coronal TIBS slices are obtained with this kV image. The image of the three SI direction rods over the sensitometry module can be seen, while the sensitometry module and other three rods almost disappeared in Fig. 3(b), the 1.3 cm thick coronal TIBS slice. Likewise, the same slice thickness of TIBS at different level is shown in Fig. 3(c), which contains the other three LR direction rods. It shows similar result of removing most of the information from the other three rods and the sensitometry module. Figure 3(d) shows a 2.4 cm TIBS slice that contains the whole sensitometry disk; all the sensitometry inserts are clearly visible (1–8 clockwise) including the one with the lowest contrast (3) that is hardly discernable on the original kV image. To allow for fair comparison, the window/level for both kV image and subtracted slice are carefully adjusted so that both images have respective optimal W/L in displaying the targeted structures. Note that all TIBS images still contained shadows from background structures; nevertheless, they provided a cleaner image of structures within the selected SOI. Figure 4 shows profiles through the images of the sensitometry disk; the profiles were drawn through the center of inserts 4 and 6 for the images of kV (lower solid), DRR_{BKG} (dashed), and their difference TIBS_{SOI} (upper solid).

The disks that were not overlapping with any shadows of background PMMA rods were selected to compute CNRs. We chose insert 1 and 5 in Fig. 3(d) and drew two same sized circles completely inside each sensitometer as ROIs. Two same sized circles were drawn in the vicinity and completely outside each sensitometer. For insert 1, the subtracted slice has CNR=7.27 compared to CNR=1.33 in the kV image, representing a 5.5-fold increase. For insert 5, the subtracted slice has CNR=28.98, compared to CNR=11.14 in the kV image, a 2.6-fold increase. The CNRs calculated from the reference images DRR_{SOI} are 18.7 and 105.9 for inserts 1 and 5, respectively.

The low-contrast inserts within phantom 2 were used to test the soft-tissue detectability with the TIBS technique. A 2.4 cm thick SOI containing inserts of different sizes and contrasts was selected and extracted from the kV image. The

FIG. 4. Plots through inserts 4 and 6 of phantom 1 for kV (lower solid), DRR_{BKG} (dashed), and their difference TIBS_{SOI} (upper solid).

TIBS image [Fig. 5(b)] showed at least seven discernable low-contrast inserts ranging from 1.5 down to 0.4 cm in diameter as pointed by the arrows. The seven visible disks correspond to inserts with contrast of 1%.²¹ The other inserts, with 0.5% or lower contrast were not visible in the TIBS image. Several inserts with 0.5% contrast²¹ can be recognized on the reference image DRR_{SOI} [on the left side of Fig. 5(c)]. 1% difference in x-ray absorption coefficient represents soft-tissue contrast range, and it is not surprising to observe none of these seven inserts on the kV image [Fig. 5(a)].

Now consider the practical case where kV imaging setup is not perfectly inline with that of CBCT scan. The kV image was intentionally shifted from its original position by ~ 1.5 mm in both SI and LR directions in the imaging plane, which mimics rigid-body setup errors prior to the application of TIBS technique. We subtracted the background DRR from the shifted kV image and obtain three slices shown in Figs. 6(a)–6(c). Compared to their corresponding ideal TIBS images [Figs. 3(b)–3(d)] it can be seen that misaligned background manifests as edge-enhanced structures ghosting on the final displayed TIBS image. It is also observed that the level of edge-ghosting artifacts proportionally depends on the size and contrast of the misaligned background

FIG. 5. Phantom 2: kV image (a), TIBS image of a 2.4 cm thick coronal slice (b), and reference image DRR_{SOI} of the same slice. Seven 1% contrast inserts ranging from 1.5 to 0.4 cm in diameter are pointed by arrows in (b); none of them is visible in (a); more inserts with 0.5% contrast can be recognized on the left of (c).

FIG. 6. Phantom 1 subtraction slices when DRR and kV images are intentionally misaligned. (a)–(c) correspond to slice locations of Figs. 3(a)–3(c), respectively. Misaligned background manifests as edge-enhanced structures ghosting on the final displayed TIBS images. A simple image registration before subtraction can eliminate these artifacts and produce images same as Figs. 3(b)–3(d).

components—i.e., the bigger size or higher contrast the misaligned components are, the heavier edge-ghosting artifacts are displayed in the subtraction.

As previously described, solution to this problem is to apply an image registration before subtracting DRR_{BKG} from kV image. In the phantom experiment where kV image is artificially translated in the imaging plane, a 2D rigid-body image registration would suffice to correct for the mismatch between kV and DRR_{BKG} . The image quality in Fig. 6 could be improved to the level comparable to that in Fig. 3 by adding a simple image registration step.

III.B. Head and neck case

A 2 cm thick sagittal SOI and a 2 cm coronal SOI were selected in a CBCT image volume of a head and neck patient, in which case the tumor resides in the sphenoid sinus. TIBS images were generated by subtracting background DRR images from the right lateral and the anterior kV images, respectively.

In this paragraph we show the results under the ideal situation, where kV images are a subset of the (week 3) CBCT data. Compared to their corresponding kV images [Figs. 7(a) and 7(d)], the TIBS images [Figs. 7(b) and 7(e)] display little background information similar to the reference images, i.e., the projections of the SOIs isolated from the CBCT image

FIG. 7. Head and neck case in LR/AP directions: kV images [(a) and (d)], TIBS images [(b) and (e)], and reference images [(c) and (f)]. Reference images are the projections of the SOIs isolated from the CBCT image as shown in Eq. (5). The target area is pointed by arrows.

FIG. 8. Series of TIBS images zoomed on the target region shown in Fig. 7(b) from different weeks of the treatment course (weeks 1–5). Increasing air volume in the sinus cavity is observed from the sequence of images. By choosing kV images from the same week CBCT datasets, all TIBS images are created assuming no setup errors.

[Figs. 7(c) and 7(f)]. The background-subtracted sagittal and coronal slices have superior image quality over the corresponding LR/AP kV images and better defined anatomical information on the SOIs. The arrows in Figs. 7(b) and 7(e) delineate the sinus cavity region where tumor resides; the tumor/air interface can be easily identified in the TIBS images while not in original kV images. A series of TIBS images zoomed on the target region is displayed in Fig. 8, where kV images were, respectively, taken from the corresponding week CBCT datasets, weeks 1–5. The air volume in the sinus cavity increased as treatments progressed, a clear indication of tumor shrinkage due to radiation therapy. Note that the images of weeks 1, 4, and 5 look noisier than those of weeks 2 and 3; this is because the CBCTs were acquired in “low-dose” mode in weeks 1, 4, and 5, which reduce the imaging dose to about one-fifth of that of the “standard-dose” mode in weeks 2 and 3.²² In both dose levels, TIBS images show improved image quality compared to their corresponding kV images.

Results considering a more clinically realistic scenario are shown below, where geometrical mismatch existed between the kV imaging and CBCT dataset. Week 2 CBCT was used as *a priori* information, while projections of week 3 CBCT were used as the LR kV images acquired on the treatment day (1 week after previous CBCT was taken). If direct subtraction was performed, TIBS images [Fig. 9(a)] contained

FIG. 9. TIBS after image registration between the kV image and *a priori* information which is the CBCT of the previous week for a head and neck case. (a) Direct subtraction, without registration. TIBS image contains edge-ghosting artifacts due to mismatched background. (b) Subtracted TIBS image, after rigid-body components of the misalignment is corrected by image registration software FSL. (c) LR direction kV image.

edge-ghosting artifacts due to mismatched background. To reduce these artifacts, we used a 2D rigid-body registration between the week 3 kV image and week 2 DRR image. The image registration considers two translation components and one rotation angle within the 2D imaging plane. The registration result from the FSL software indicated a 1.57 mm AP direction shift and a 1.97 mm SI direction shift on the imaging plane and nearly no rotation between week 3 kV image and week 2 DRR image. This corresponds to 1.04 and 1.3 mm isocenter shift in the AP and SI directions, respectively, between the two treatment sessions. After this translational mismatch was accounted for, TIBS image [Fig. 9(b)] provides considerably better image quality compared to direct subtraction result [Fig. 9(a)], and it provides more accurate SOI than the conventional kV image [Fig. 9(c)].

IV. DISCUSSIONS

It is usually not possible to delineate the target on conventional kV images directly because it may be obscured and overshadowed by background structures. CBCT scans acquired at the time of treatment with gantry-mounted kV imaging equipment provide images with good, albeit still not optimal, soft-tissue contrast. CBCT requires at least 2 min of extra operating time and gives the patient a dose between 1.5 and 6 cGy,¹⁰ depending on the site and technique. The CBCT imaging dose often hampers its frequent uses, especially in head and neck cases.

We proposed a novel TIBS technique that extracts slice of interest from a kV image by removing information of background slices. The method utilizes the *a priori* anatomical information on the patient that is usually available in previously acquired CT datasets, e.g., planning fan beam CT or cone beam CT. In this work, CBCT is used as the source of *a priori* information to match the attenuation coefficients to those of the kV image. However, other types of CT are not necessarily excluded provided there is a good attenuation mapping between two x-ray techniques with different energy spectra.

Two main premises are needed in order for our TIBS technique to produce the ideal slice image, i.e., the projection of the linear attenuation coefficients of a selected slice of the anatomy. First, the *a priori* information should be accurate, such that the effective x-ray attenuation map of the scan object is obtained as accurately as possible. Sources of inaccuracy that typically affect the image quality of the CBCT dataset we used include (1) cone beam artifacts, especially severe for large cone angles in circular CBCT, and (2) motion related artifacts that are inevitably present for CBCT scans of thorax or abdomen regions due to patient breathing. Our first premise is approximately valid for small cone angles and when patient motion is negligible, such is the case for the head and neck case used in this paper. In the second place, the *a priori* information should be perfectly utilized, such that the anatomy at the time of kV imaging matches to that of CBCT imaging. The following conditions are assumed for the second premise: (1) Changes in patient's anatomy between the time of kV imaging and the CBCT

imaging do not occur outside the selected SOI and (2) CBCT and kV imaging setups are geometrically matched. The first assumption means major target motions and/or deformations are contained within the SOI or changes occur in the background are negligible. This is reasonable for most brain, head and neck, and some pelvic cases since the anatomy does not move or morph significantly with the exception of motions due to respiration or severe weight loss. The second assumption is relatively strict due to the nature of the subtraction methodology. Any slight geometric mismatches will be magnified in the final subtracted slice, causing increased structured noise and depriving the clinical value of subtracted slice. An adequate image registration step is required to align the background components of the CBCT and kV images as close as possible. The pre-generated reference image DRR_{SOI} can be used to identify the usability of TIBS and to regenerate or discard the TIBS image if unsuccessfully corrected misalignments result in unacceptable image quality. TIBS provides a way to improve the accuracy of image guidance using x-ray projections in tasks such as daily patient positioning or tumor regression monitoring as described in this paper. With the aid of a simple rigid-body registration step, the feasibility and capability of TIBS scheme have been demonstrated. In order for the method to be applied clinically, more sophisticated image registration algorithms are required in order to accurately align the background of previous CBCT and current kV images, including deformable components. This is in our future research interest.

TIBS does not require special hardware other than what is used in conventional kV and CT imaging. Quantitative utilization of previous CT data for the purpose of enhancing the image quality of x-ray radiographs is a novel concept, and it could find unique applications within the context of image-guided radiation therapy. Requiring only snapshot radiographs, TIBS allows for real-time tumor localization without implanted markers. This method might be potentially extended to x-ray fluoroscopy and megavoltage x-ray imaging and also to intrafractional image guidance.

V. CONCLUSIONS

In this paper, we propose a novel imaging technique (TIBS) that subtracts the background information out of a kV image, yielding an image of slice of interest only. Results with phantoms and a clinical case show that the TIBS technique effectively reduces ambiguities and enhances contrast over conventional kV imaging. The TIBS technique can provide excellent contrast for a section of an object with snapshot x-ray projections, therefore reducing the need for frequent uses of CT scans that requires excessive imaging time and dose for image guided radiation treatments.

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