A novel technique for markerless, self-sorted 4D-CBCT: Feasibility study

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Purpose: Four-dimensional CBCT (4D-CBCT) imaging in the treatment room can provide verification of moving targets, facilitating the potential for margin reduction and consequent dose escalation. Reconstruction of 4D-CBCT images requires correlation of respiratory phase with projection acquisition, which is often achieved with external surrogate measures of respiration. However, external measures may not be a direct representation of the motion of the internal anatomy and it is therefore the aim of this work to develop a novel technique for markerless, self-sorted 4D-CBCT reconstruction.

Methods: A novel 4D-CBCT reconstruction technique based on the principles of Fourier transform (FT) theory was investigated for markerless extraction of respiratory phase directly from projection data. In this FT technique, both phase information (FT-phase) and magnitude information (FT-magnitude) were separately implemented in order to discern projections corresponding to peak inspiration, which then facilitated the proceeding sort and bin processes involved in retrospective 4D image reconstruction. In order to quantitatively evaluate the accuracy of the Fourier methods, peak-inspiration projections identified each by FT-phase and FT-magnitude were compared to those manually identified by visual tracking of structures. The average phase difference as assigned by each method vs the manual technique was calculated per projection dataset. The percentage of projections that were assigned within 10% phase of each other was also computed. Both Fourier methods were tested on two phantom datasets, programmed to exhibit sinusoidal respiratory cycles of 2.0 cm in amplitude with respiratory cycle lengths of 3 and 6 s, respectively. Additionally, three sets of patient projections were studied. All of the data were previously acquired at slow-gantry speeds ranging between 0.6°/s and 0.7°/s over a 200° rotation. Ten phase bins with 10% phase windows were selected for 4D-CBCT reconstruction of one phantom and one patient case for visual and quantitative comparison. Line profiles were plotted for the 0% and 50% phase images as reconstructed by the manual technique and each of the Fourier methods.

Results: As compared with the manual technique, the FT-phase method resulted in average phase differences of 1.8% for the phantom with the 3 s respiratory cycle, 3.9% for the phantom with the 6 s respiratory cycle, 2.9% for patient 1, 5.0% for patient 2, and 3.8% for patient 3. For the FT-magnitude method, these numbers were 2.1%, 4.0%, 2.9%, 5.3%, and 3.5%, respectively. The percentage of projections that were assigned within 10% phase by the FT-phase method as compared to the manual technique for the five datasets were 100.0%, 100.0%, 97.6%, 93.4%, and 94.1%, respectively, whereas for the FT-magnitude method these percentages were 98.1%, 92.3%, 98.7%, 87.3%, and 95.7%. Reconstructed 4D phase images for both the phantom and patient case were visually and quantitatively equivalent between each of the Fourier methods vs the manual technique.

Conclusions: A novel technique employing the basics of Fourier transform theory was investigated and demonstrated to be feasible in achieving markerless, self-sorted 4D-CBCT reconstruction. © 2012 American Association of Physicists in Medicine. [DOI: 10.1118/1.3685443]

Key words: 4D cone-beam CT, localization, on-board imaging, organ motion, image-guided radiotherapy

I. INTRODUCTION

According to the American Cancer Society, carcinoma of the lung and bronchus continued to occupy the leading role in cancer deaths in 2010, an estimated 29% among men and 26% among women in the United States.¹ These carcinomas are often inoperable due to compromised lung functions and therefore are recommended for radiotherapy, where they present a challenge to the entire treatment process because

they are subject to respiratory-induced motion. There has thus been substantial investigation and development of motion-management strategies in the field of radiation therapy. Motion management is crucial because it can potentially result in margin reduction, which may allow for dose escalation and consequently improved local control. For example, four-dimensional CT (4DCT) has highly evolved since it was first introduced in 2003 and has now become the standard of care during the simulation process of tumors

affected by respiratory motion, especially because it has been widely reported that this motion is very patient-specific.^{2–19} 4DCT provides volumetric information of the target as it moves during respiration by rapidly acquiring axial CT slices for at least one respiratory cycle per couch position and then retrospectively sorting the reconstructed slices according to phase in order to form a static 3D volume per window of respiration.^{20–25}

Recently, 4D cone-beam CT (4D-CBCT) imaging has been proposed for motion management during the patient localization stage of radiotherapy with the use of the (kV or MV) on-board imager that is mounted to the gantry of the treatment machine.^{26–32} 4D-CBCT images are generated quite differently from 4DCT because the acquisition time is limited by the rotational speed of the gantry. Instead of combining reconstructed axial slices per respiratory phase, 2D projections are acquired over multiple respiratory cycles and then sorted according to phase prior to reconstruction. Therefore, 4D-CBCT also results in time-resolved volumetric images of the target as it moves during a patient's respiration.

In order to generate 4DCT/4D-CBCT datasets, the acquired slice/projection must be correlated with respiratory phase. Phase is defined as the percentage of respiration that has transpired since peak-inspiration (usually denoted as 0%). Typically, this information is acquired with the use of external markers that serve as surrogates for the measurement of respiratory phase, by detecting changes in abdominal displacement or lung air volume. Commercially available products for detecting abdominal displacement include the Varian Real-time Position Management (RPM) system (Varian Medical Systems, Palo Alto, CA) (Refs. 21, 22, 24, 25, 33, and 34) and the strain-gauged Anzai belt (Anzai Medical Systems, Tokyo, Japan).^{35,36} The Varian RPM system consists of a camera and an infrared reflecting marker, whereas the Anzai belt is composed of an elastic belt with a load cell that detects changes in pressure as a patient breathes. To detect changes in lung air volume, spirometry has been implemented to serve as an external surrogate for respiratory phase as well.³⁷ All of these external markers are assumed to correlate with the displacement of internal anatomy, but studies have demonstrated that uncertainties exist in this relationship.^{38–41} Mismatching of projections and respiratory phase can lead to artifacts in the reconstructed phase images and thus may potentially lead to errors in target volume delineation. Ideally, the respiratory phase would be derived directly from the respiratory-induced motion of the anatomy, without reliance on external surrogate measures.

As a result of the limitations of external markers, there have been several techniques proposed for determining respiratory phase from CBCT projection images without the use of external markers. One such technique is known as the Amsterdam shroud technique, proposed by Zijp *et al.*,⁴² and consists of enhancing the diaphragm in individual projections by: (1) applying a logarithm, (2) performing edge-detection and thresholds, (3) collapsing each projection onto the craniocaudal axis, and (4) successively combining the

1D line images into a 2D image, from which the displacement of the diaphragm is finally extracted by aligning each 1D line image to the next. However, this technique relies on a visible oscillating structure to be present across all projection angles, and so Kavanagh *et al.*⁴³ proposed a modified version of this technique that sums all pixels in the vertical and horizontal direction in a region encompassing the lungs and then plots this value across the acquired projections. The signal is then further corrected for rotational changes of the source and detector. The resultant plot provides the respiratory signal (even for datasets unsuitable for the Amsterdam shroud technique), to which phase-based sorting can be applied for 4D-CBCT reconstruction. The robustness of this technique is unclear as it was only compared with the Amsterdam shroud technique on fluoroscopy images and further patient CBCT data were not presented.

Other ways of deriving motion from cone-beam projections for 4D reconstructions were studied by Rit *et al.*,⁴⁴ Siochi,⁴⁵ Lewis *et al.*,⁴⁶ and Park *et al.*⁴⁷ Rit *et al.* developed a methodology that selects thousands of points of interests in the projections, then follows each of these points across projections with a Block Matching Algorithm, and finally performs some signal processing on the acquired trajectory.⁴⁴ However, their results showed that the accuracy of the sorting of projections depended on the number of desired phase bins. Siochi developed a different technique for MV cone-beam projections by determining a bounding box for diaphragm motion (for all projections) based on two pairs of full-inhale and full-exhale views.⁴⁵ This technique was tested with a moving tungsten pin and several lung patients, but it is important to note that it relies on the diaphragm being present in the field of view. Lewis *et al.* present a direct tumor tracking algorithm that takes the 4DCT reference volume at each phase, registers these to the CBCT based on bony anatomy, generates DRRs at all angles corresponding to a CBCT acquisition, and then uses normalized mutual information to find the best match between DRR and CBCT projections to locate the tumor in 2D, from which a 3D tumor trajectory is constructed with minor approximations.⁴⁶ This algorithm worked quite well for the physical and digital phantom cases presented, but did run into some trouble for the two patient cases because of the presence of other high contrast objects blocking the view of the tumor in the projection. Lastly, Park *et al.* also implemented a template matching technique to extract respiratory signals from cone-beam projections that was applied to a phantom and five patient cases for 4D-CBCT and 4D-DTS reconstruction. It is important to note that their technique is restricted to liver tumors and relies on the insertion of metal fiducial markers.⁴⁷

All of the aforementioned techniques selected specific anatomical information as input for signal analysis and each had its own benefits and limitations in deriving respiratory signals from projection data without the use of external surrogates. Recently, Cai *et al.*⁴⁸ presented preliminary results on using the Fourier transform (FT) to extract respiratory signals from continuously acquired magnetic resonance (MR) images for the reconstruction of 4D-MRI. The work presented here forth is a feasibility study in which the

concepts of the FT are similarly applied to cone-beam projection data in order to facilitate the correlation between respiratory phase and projection data. The major advantage of using a FT technique is that both phase and magnitude information are available for extraction respiratory signals. The proposed technique further enhances the available methodologies by providing a much simpler and more efficient method for markerless, self-sorted 4D-CBCT reconstruction than what has previously been implemented. The novelty of the technique is that it characterizes the anatomical motion information directly from kV projections by tracking the changes in key FT parameters of continuously acquired projection images. The technique simplifies the 4D-CBCT reconstruction process by eliminating the need for additional measurements of respiratory phase.

II. METHODS AND MATERIALS

II.A. Fourier phase vs magnitude technique

The motivation for this novel technique comes from the basics of FT theory. The FT of an object is often described by its magnitude and phase, which are all mathematically described in the equations below with $Re(F)$ signifying the real component of $F(u,v)$ and $Im(F)$ signifying the imaginary component of $F(u,v)$

$$F(u, v) = \int_{-\infty}^{\infty} f(x, y) e^{-i2\pi(ux+vy)} dx dy, \quad (1)$$

$$|F(u, v)| = \sqrt{Re(F)^2 + Im(F)^2}, \quad (2)$$

$$\varphi = \arctan\left(\frac{Im(F)}{Re(F)}\right). \quad (3)$$

The magnitude of the FT of an image represents how much of a particular frequency component is present in that image, whereas the phase represents where in the image that particular frequency component is present.

An important property of the phase information of a FT is demonstrated with the shift theorem, which dictates that a geometric shift induced in Cartesian space corresponds to a phase shift in Fourier space.⁴⁹

$$\begin{aligned} \int_{-\infty}^{\infty} f(x-a) e^{-i2\pi xs} dx &= \int_{-\infty}^{\infty} f(x-a) e^{-i2\pi(x-a)s} e^{-i2\pi xs} d(x-a) \\ &= e^{-i2\pi as} F(s). \end{aligned} \quad (4)$$

Because the goal is to extract motion information from projection data and motion presents in the projections as consecutive shifts in Cartesian space, then it is reasonable to hypothesize that these changes in Cartesian space (i.e., motion) would correspond to detectable changes in phase in Fourier space.

Alternatively, the magnitude information of the FT can also be used for respiratory phase extraction. During image acquisition, organ motion presents itself in the projection data as changes in intensities in two different manners:

(1) the diaphragm moving up and down and (2) the thorax expanding and retracting, both of which are a result of respiration. These changes in intensity can be related to changes in the magnitude component of the 2D FT, thus establishing another potential relationship between the motion contained in the projections over time and their corresponding Fourier magnitude information.

Both of the proposed methods initially involve the same set of image processing procedures. Once the projections are acquired, the logarithm of each projection is taken. This enhances the contrast because the dynamic range of the images is compressed, minimizing the large variation in pixel values that is typically seen in projection data. The next step involves selecting a region-of-interest (ROI), simply starting at the superior portion of the lungs and including everything below. The entire width of the projection data is maintained. Because the amount of lung in the FOV can vary across patients, the ROI is then symmetrically padded with zeros to restore the data to its original projection size. This allows us to later select the same frequency with the FT-phase technique for all patients, independent of the ROI size. The last step that both methods have in common is the calculation of the 2D Fourier transform (2D FT) of each projection in the dataset.

The FT-phase technique then requires computation of the phase at every pixel. The first low frequency location along the y-axis (0,1) is then selected and this phase value is plotted as a function of every projection in the dataset. Low frequency is chosen because the overall motion of structures (i.e., the respiratory motion) is mainly encoded as low frequency information in Fourier space. The y-axis is relevant because it represents motion occurring along the superior-inferior direction. It is important to note that slow-gantry acquired projection data works best with this technique since it is sensitive to changes between projections and ideally the phase shifts would only come from respiratory-induced motion rather than changes in anatomy due to rotation of the source around the patient. The plotted phase values constitute the extracted signal, which has a low frequency component (due to varying attenuation of the photons around the patient), a high frequency component (due to respiration) that is particularly of interest, and an even higher frequency component (due to cardiac motion). Some minor signal processing is required to separate the component of interest from the other two, after the signal is “unwrapped” in order to minimize the jumps that may occur in phase between consecutive data points by adding multiples of $\pm 2\pi$. In order to remove the cardiac motion, a low-pass filter is first applied. To detrend the signal from the low frequency component, the signal is heavily smoothed with a large moving average, which is then subtracted from the original signal. Some additional smoothing with a much smaller moving average is applied to facilitate automatic extraction of the valleys of the FT-phase signal, which evidently correspond to peak-inspiration projections. All image processing and data analysis was performed with an in-house developed MATLAB program (The Mathworks, Inc. Natick, Massachusetts).

The FT-magnitude technique requires a slightly different procedure. First, the magnitude (instead of phase) is calculated at every pixel. Here, the DC component is selected, rather than a low frequency magnitude. Although, other low frequency locations may work, the DC component was found to be more robust (see Sec. III). The DC component is of interest because it is essentially the average of the intensity values in the projections, which conveniently change as a result of the motion of the anatomy that is due to respiration and rotation of the x-ray source. Thus, the DC component is plotted across the projections in the dataset to form the FT-magnitude extracted signal. The same aforementioned signal processing is then applied to this signal as well. The only difference is that for this technique, the peaks of the extracted signal correspond to peak-inspiration projections. The procedure for both techniques is summarized in Fig. 1.

II.B. Phantom and patient projection data

The proposed Fourier methods were both tested with phantom and patient data that were previously acquired under an investigational 4D-CBCT study.³² Both phantom and patient projections were acquired with an x-ray flat panel detector mounted orthogonally to the MV treatment gantry on the Varian Trilogy system (Medical Systems, Palo Alto, CA). The phantom imaged was an International Electrotechnical Commission (IEC) body phantom with six hollow acrylic spheres of various diameters: 1.1, 1.4, 1.8, 2.2, and 2.7 cm. The IEC phantom was placed on a moving platform, which was programmed to simulate regular respiratory cycles that were 3 and 6 s in duration. The motion was limited to the superior-inferior direction with an amplitude of 2.0 cm. Both respiratory cycles were imaged at a slow-gantry speed resulting in a scan length of 5 min over a

200° arc. The 3 s respiratory cycle resulted in 1402 projections and the 6 s respiratory cycle had 1405 projections.

In addition to the two phantom data sets of projections, the proposed algorithms were also tested with three sets of patient projections acquired over a 200° arc at slow-gantry speeds of approximately 0.71°/s, 0.70°/s, and 0.60°/s. The resulting scan lengths were 4.5, 4.5, and 5.7 minutes acquired with frame rates of 7, 7, and 5 fps, respectively. Patient 1 had a total of 1982 projections, patient 2 had 2005 projections, and patient 3 had 1679 projections. Further details on the CBCT acquisition parameters for this data can be found from Lu *et al.*³²

II.C. Evaluation

In order to investigate how accurately the proposed Fourier methods (phase and magnitude) were identifying the peak-inspiration phase (0%), the respiratory signal was also derived by visually tracking structures in the data set. Projection numbers that corresponded to peak inspiration were recorded, and this served as the gold standard for comparison.

Quantitative comparison between methods was implemented by computing respiratory phase, defined as the percentage of the respiratory cycle that has passed since peak inspiration (represented by 0% or 100%). Upon determining peak-inspiration, every projection was then assigned a phase according to linear interpolation in between consecutive peak-inspiration projections. This was performed for both Fourier methods as well as the manual technique. The average difference in phase across the entire dataset for FT-phase vs manual and FT-magnitude vs manual was then computed

$$\text{Average difference (\%)} = \frac{1}{n} \sum_{i=1}^n |\text{Phase}(i_{\text{Man}}) - \text{Phase}(i_{\text{FT}})|. \quad (5)$$

The percentage of projections that were assigned within 10% phase of each other when comparing each of the two Fourier methods to the manual technique was also computed per dataset.

$$\text{Percentage (\%)} = 100\% * \frac{N_{\text{Projs}} \{ |\text{Phase}(i_{\text{Man}}) - \text{Phase}(i_{\text{FT}})| \leq 10\% \}}{\text{Total Projs}}. \quad (6)$$

The purpose of this computation was to get an idea of how accurately the projections would still be sorted upon phase assignment according to each of the two Fourier methods. A 10% phase criterion was chosen because this is the conventional amount of motion present in each phase bin when reconstructing 4D images. Both of the quantitative evaluations were performed on all five projection datasets for the FT-phase and FT-magnitude methods.

II.D. ROI selection

It was important to investigate if the accuracy of one particular method relied on what region of the projection was

FIG. 1. Summary of procedures for the FT-phase vs FT-magnitude techniques.

TABLE I. Results for the average difference in phase using the entire projection vs ROI for both Fourier techniques relative to the manual one.

Projections	Entire projection: Average difference in phase (%)		ROI: Average difference in phase (%)	
	FT—Phase	FT—Magnitude	FT—Phase	FT—Magnitude
Phantom RC = 3 s	1.8	7.4	1.8	2.1
Phantom RC = 6 s	3.7	9.3	3.9	4.0
Patient 1	5.1	2.9	2.9	2.9
Patient 2	15.5	6.8	5.0	5.3
Patient 3	6.7	3.4	3.8	3.5

input into the Fourier transform. This is relevant because different areas of the projection experience different amounts of intensity changes depending on the magnitude of organ displacement. In the interest of robustness and simplicity, the selected ROI began at the top of the lungs and proceeded all the way downward in the vertical direction, meanwhile including all of the projection information in the horizontal direction (to account for the changing location of the lungs in the projection data as a function of x-ray source angle). The quantitative comparisons presented in Sec. II C were performed with using the entire projection as input vs the selected ROI for all five projection datasets (Note: for the IEC phantom data the ROI consisted of the moving portion of the rods). This was done in order to assess whether selecting an ROI is a crucial factor in determining the accuracy of the technique.

II.E. Frequency selection

As previously mentioned, low frequency locations in Fourier space are of interest because large anatomical motion is typically represented as low frequency information. It was thus important to assess which low frequency location was most robust across the accessible projection datasets. Both quantitative metrics of average phase difference and percentage of projections assigned within 10% phase were again computed across low frequency locations ranging from (0, 0) to (5, 5). Based on these results, the appropriate low frequency location was recommended for selection for each of the Fourier techniques.

II.F. 4D-CBCT reconstruction

In order to provide some additional results for how the different methods measured against each other, 4D-CBCT phase images were reconstructed according to the sorting

and binning that occurred as a result of the extracted peak-inspiration projections for each method. The sorting process can be performed either according to phase or amplitude of the respiratory signal. However, the resultant signals from both of the proposed Fourier methods exhibit baseline shift and thus must be sorted according to phase, rather than amplitude. Therefore, once the peak-inspiration projections were identified, the projections in between adjacent peak-inspirations (0%) were assigned phases based on linear interpolation. The projections were then appropriately sorted into 10 phase bins, each with 10% phase windows, which is the typical convention among current investigators of 4D imaging. Each bin was centered at 0%, 10%, 20%, etc., and contained the adjacent 5% phases within the bin so that the 10% bin really corresponded to phases between 5% and 15%.

Upon completion of the sort and bin procedure, each of the ten phase bins was then reconstructed with an in-house Feldkamp-type reconstruction algorithm.^{50,51} Each reconstructed volume was of size $512 \times 320 \times 401$ with 0.5 mm slice thickness and 0.507 mm^2 in-plane pixel size. This was performed for one phantom (RC = 6 s) and one patient (patient 1) case across all three methods (manual, FT-phase, FT-magnitude). In order to additionally assess the images quantitatively, line profiles were plotted to compare edge preservation for two of the ten phase bins as reconstructed by the FT-phase and FT-magnitude method relative to the manual technique.

III. RESULTS

Tables I and II demonstrate the effects of selecting the ROI vs using the entire projection on the average phase difference and the percentage of projections assigned within 10% phase of the manual technique. Both Fourier techniques are evaluated with these quantitative metrics across all

TABLE II. Inputting the entire projection vs ROI results for the percentage of projections that were within 10% phase of each other for each Fourier technique as compared with the manual technique.

Projections	Entire projection: Percentage of projections within 10% phase (%)		ROI: Percentage of projections within 10% phase (%)	
	FT—phase	FT—magnitude	FT—phase	FT—magnitude
Phantom RC = 3 s	100.0	86.3	100.0	98.1
Phantom RC = 6 s	100.0	79.9	100.0	92.3
Patient 1	92.5	98.7	97.6	98.7
Patient 2	58.3	83.1	93.4	87.3
Patient 3	83.2	96.1	94.1	95.7

FIG. 2. The first row of bar graphs demonstrates the average phase difference as a function of low frequency location along the y-axis for both Fourier techniques. The second row shows the same results for a range of low frequency locations in both the x and y dimensions.

projection datasets. It is evident that the ROI selection does play a crucial role in determining the accuracy of the FT-phase technique, where it leads to a much lower average phase difference and much greater percentage of projections assigned within 10% phase for the patient data. The effect of the ROI selection is less dramatic for the DC component of the FT-magnitude technique.

Figure 2 displays the average phase difference as a result of varying low frequency locations for both the FT-phase and FT-magnitude methods. The top row emphasizes the differences solely along the y-dimension from (0,0) to (0,5). Note that the (0,0) value is not plotted for the FT-phase technique because the phase angle at that location is always 0. The bottom row demonstrates the differences along both x and y dimensions for a range of values from (0,0) to (5,5). Intuitively, it is evident that the lowest average phase difference occurs mainly for low frequencies located along the y-dimension for both Fourier techniques.

Further, the graphs depict that the FT-phase technique exhibits the lowest average phase difference at the first low frequency location (0,1) when comparing across all patient datasets. However, for the FT-magnitude technique the most robust value is clearly the DC component for the three studied patients.

Figure 3 studies the same effect, but measures it with the more meaningful quantitative metric of percentage of projections assigned within 10% phase of the manual technique. Similarly, it is obvious that the highest bars are seen for frequencies along the y-axis (superior-inferior direction of motion) for both methods. Taking into account all three patients, the first low frequency location of (0,1) again gives the best results for the FT-phase technique. In accordance with the average phase difference results, the DC component of the FT-magnitude technique consistently leads to the largest percentage of projections assigned within 10% phase for the three patient cases.

FIG. 3. Percentage of projections assigned within 10% phase plotted as a function of FT-phase and FT-magnitude techniques for a variety of low frequency locations. The top row emphasizes the results along the y-dimension, meanwhile the bottom row of graphs displays the results along both dimensions.

FIG. 4. Extracted signal from both Fourier techniques for patient 1. Peak-inspiration determined from manual identification is marked with circles and peak-inspiration determined from the Fourier technique is marked with squares.

Figures 4–6 present the extracted signals according to both Fourier techniques for patients 1–3. The red circles on the plots represent the gold standard (manual) identified peak-inspiration projections and the black squares represent the automatically extracted ones for the corresponding listed technique. For the most part, it is evident that the circles and squares match up nicely, further supporting the results seen in Tables I and II. Both Fourier techniques are adequate for 4D-CBCT reconstruction in that they result in low average phase differences (5.0% and less) and a large majority (87% and greater) of projections assigned within 10% phase of the gold standard. Generally, both Fourier techniques give similar results, with the exception of patient 2, where the FT-phase technique shows a slight advantage (approximately 6%) over FT-magnitude in terms of percentage of projections assigned within 10% phase.

Figures 7 and 8 each display 4D-CBCT reconstructed images of five selected phase bins (0%, 20%, 50%, 70%, and 90%) of the phantom with a 6 s respiratory cycle ($RC = 6$ s) and of patient 1, respectively. Both figures present the 4D reconstructed phase images and their associated line profiles along the vertical dimension (for peak-inspiration and peak-expiration) as a function of each method: Manual vs

FT-phase vs FT-magnitude. As expected from the results of the quantitative metric comparisons of the phantom with $RC = 6$ s, no visually detectable differences are evident across all three techniques per selected phase bin in Fig. 7. The agreement between the plotted line profiles for the 0% and 50% phase bins also demonstrate that all three methods result in equivalent 4D-CBCT reconstructions. There are also no visually identifiable differences present for the 4D-CBCT images reconstructed for patient 1 in Fig. 8. Although not as ideal as the phantom data, the line profiles do display nice agreement and thus further validate the equivalence of the two Fourier methods to the gold standard.

IV. DISCUSSION

The results presented in this article have demonstrated the preliminary implementation of a novel technique based on 2D image FT for markerless and self-sorted 4D-CBCT reconstruction of images. In general, both of the proposed Fourier methods accomplished the 4D sort and bin processes equivalently. Table I (ROI) depicts that both techniques resulted in a low average phase difference (around 5% and below). More importantly, Table II (ROI) demonstrates that

FIG. 5. FT-phase vs FT-magnitude derived signal for patient 2. Circles indicate peak-inspiration identified with the reference technique whereas squares indicate peak-inspiration identified by the displayed technique.

FIG. 6. The plotted signal from each FT method for patient 3. Peak-inspiration determined from visual identification is displayed with circles and from automatic Fourier identification with squares.

FIG. 7. 4D-CBCT reconstructed phase images of the IEC phantom exhibiting 6 s respiration for the reference technique vs FT-phase technique vs FT-magnitude technique. Line profiles are plotted for 0% and 50% phase bins.

FIG. 8. Five selected phase reconstructions of CBCT images of patient 1 for visual assessment of any differences between the two proposed Fourier techniques to the gold standard. Plotted line profiles are displayed for the peak-inspiration and peak-expiration phases.

both techniques were able to sort a large majority (around 87% and above) of the projections within 10% phase of the gold standard method of manually identifying peak-inspiration projections. This is promising because it leads to phase differences that are within the acceptable window (10%) of motion that each reconstructed phase image typically has. Further, this novel technique does not require the presence of the diaphragm in the FOV, as we can see from the successful results of patients 2 and 3, which are imaged above the diaphragm.

For this feasibility study, it has been demonstrated that the region of the projection selected as input into the algorithm influences the results. The results in Tables I and II revealed that ROI selection starting with the lungs and including all anatomy below improved the accuracy of peak-inspiration identification. This type of straightforward ROI selection can easily be automated with some minor additional image processing and is therefore not an obstacle for the clinical translation of this methodology. For the three patients studied, the size of the ROI also did not seem to hamper the robustness of the frequency location selected for the FT-phase technique, as long as the ROIs were padded with zeros to restore the original projection size.

The results presented in Figs. 2 and 3 may lead us to believe otherwise for the FT-magnitude technique. From the bar graphs of the quantitative metrics at different frequency locations, it is obvious that the DC component is the most robust for these three patients in terms of delivering the best accuracy. However, looking closely at the frequency locations along the y-axis it is also evident that for different patients there are other low frequency locations that are able to deliver just as accurate results. For example, patient 1 achieves equally good results as the DC component at locations (0,3) and (0,4), whereas patients 2 and 3 achieve equally good results at (0,1). This is likely related to the ROI size, as patient 1 had a larger proportion of the lungs in the FOV than patients 2 and 3. Padding with zeros does not seem to account for the variable ROI size for the FT-magnitude technique as it is more sensitive to the amount of a particular frequency component present in the image. Reducing the ROI size of patient 1 to match that of patient 2 shifted the good results from (0,3) and (0,4) up to (0,1) and (0,2) as hypothesized.

As there were only three sets of patient data examined in this feasibility study, it is apparent that the accuracy and robustness of the technique needs further assessment. Ideally, more slow-gantry patient data would allow for stronger validation of the technique. However, the implementation of slow-gantry CBCT acquisition of patients has been limited and so an alternative way to obtain this data is with the XCAT digital phantom that has been developed by Segars *et al.*^{52,53} This is currently the subject of ongoing investigation. It is also important to note that the acquisition parameters of the slow-gantry data used to test the Fourier techniques can be improved in terms of time and dose, while still satisfying the sampling conditions required for 4D-CBCT reconstruction, as proposed by Maurer *et al.*⁵⁴ Their work suggests that the image acquisition parameters can be

optimized per patient respiratory cycle, while maintaining a reasonable imaging dose and scan time, as well as achieving adequate 4D phase images. Employing their technique to simulate appropriate gantry rotation speeds and frame rates for specific respiratory cycles with the XCAT phantom is one of the central factors in our ongoing investigations into the robustness and accuracy of the technique.

V. CONCLUSIONS

The feasibility of using the Fourier transform technique for markerless, self-sorted 4D-CBCT reconstruction has effectively been shown. Both Fourier phase and magnitude methods resulted in low average phase differences and a high percentage of projections assigned within 10% phase of the gold standard method for all datasets. Furthermore, the 4D phase image reconstructions of the FT-magnitude technique and the FT-phase technique were qualitatively and quantitatively equivalent to the gold standard technique. Overall, the results suggest that the proposed techniques are promising and may eliminate the need for external markers in the 4D-CBCT reconstruction process.

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