High energy x-ray phase contrast CT using glancing-angle grating interferometers

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Purpose: The authors present initial progress toward a clinically compatible x-ray phase contrast CT system, using glancing-angle x-ray grating interferometry to provide high contrast soft tissue images at estimated by computer simulation dose levels comparable to conventional absorption based CT.

Methods: DPC-CT scans of a joint phantom and of soft tissues were performed in order to answer several important questions from a clinical setup point of view. A comparison between high and low fringe visibility systems is presented. The standard phase stepping method was compared with sliding window interlaced scanning. Using estimated dose values obtained with a Monte-Carlo code the authors studied the dependence of the phase image contrast on exposure time and dose.

Results: Using a glancing angle interferometer at high x-ray energy (~ 45 keV mean value) in combination with a conventional x-ray tube the authors achieved fringe visibility values of nearly 50%, never reported before. High fringe visibility is shown to be an indispensable parameter for a potential clinical scanner. Sliding window interlaced scanning proved to have higher SNRs and CNRs in a region of interest and to also be a crucial part of a low dose CT system. DPC-CT images of a soft tissue phantom at exposures in the range typical for absorption based CT of musculoskeletal extremities were obtained. Assuming a human knee as the CT target, good soft tissue phase contrast could be obtained at an estimated absorbed dose level around 8 mGy, similar to conventional CT.

Conclusions: DPC-CT with glancing-angle interferometers provides improved soft tissue contrast over absorption CT even at clinically compatible dose levels (estimated by a Monte-Carlo computer simulation). Further steps in image processing, data reconstruction, and spectral matching could make the technique fully clinically compatible. Nevertheless, due to its increased scan time and complexity the technique should be thought of not as replacing, but as complimentary to conventional CT, to be used in specific applications. © 2014 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4860275>]

Key words: x-ray phase contrast ct, glancing angle grating interferometer, high fringe visibility

1. INTRODUCTION

Differential x-ray phase contrast CT with Talbot-Lau grating interferometers (DPC-CT) has the potential to combine the high soft tissue contrast comparable to MRI with the high spatial resolution and penetration depth of x-rays. DPC-CT has proven ability to work with conventional x-ray tubes¹⁻³ and has shown promise for clinical applications.⁴⁻⁹

The technique works by placing three microperiodic gratings between the x-ray tube and the detector, creating a so

called Talbot-Lau interferometer, which converts the micro-radian deviations of x-rays caused by refraction in the sample into intensity modulations detectable by a conventional detector. Two of the gratings are absorption gratings made typically of gold, and the third is a transmission (“phase”) grating made of silicon or nickel. During the CT scan one of the gratings is laterally moved in steps of a fraction of a period (“phase-scanning”), making it possible to separate the absorption, phase-gradient (refraction angle), and ultra-small angle scattering (“dark field”) components of the image, thus

providing three different characteristics of the same sample simultaneously.^{2,10}

In the recent years there has been a lot of interest in the DPC-CT technique from the medical community in the hope for improved cancer detectability in, for instance, the breast,⁴ liver,⁵ brain,⁶ or pancreas,⁷ and for improved imaging of cartilage,⁸ or of plaque.⁹ While the experiments, both using conventional tubes and synchrotron x-rays, demonstrated improved soft tissue contrast, the dose delivered to the sample (and eventually to a patient) that is necessary to obtain this additional information still remains a big concern.¹¹ Published experiments have so far routinely exceeded the dose limits acceptable in conventional CT by up to a few orders of magnitude.^{12–14} The question yet to be answered is thus whether x-ray phase contrast imaging using grating interferometer will still outshine absorption based imaging when used with a considerably lower dose, closer to clinical acceptable values.

The optical performance of the Talbot-Lau interferometer is described by two main parameters: the interferometric fringe visibility (V), also known simply as visibility, and the angular sensitivity (S). The visibility characterizes the amplitude or contrast of the intensity modulation in the interference pattern and is defined as $V = [(I_{\max} - I_{\min}) / (I_{\max} + I_{\min})] \cdot 100\%$, where $I_{\max}$ and $I_{\min}$ are, respectively, the minimum and maximum values of the intensity pattern produced by the phase grating in each detector pixel. The angular sensitivity defines the smallest refraction angle detectable by the system. Theoretical analysis predicts that the ratio between the NEQ (noise equivalent quanta) in DPC to that in absorption CT increases proportionally to the product $S^2 \cdot V^2$.¹⁵ The angular sensitivity is determined by the ratio between the distance between G1-G2 and the G1 period, divided by the magnification of the setup. Thus, it is practically limited by the feasible grating period ($\geq \text{few } \mu\text{m}$) and by the clinically compatible scanner length (≤ 1.5 m approximately). Thus, one of the most promising ways to improve image quality at a given dose is therefore to increase the fringe visibility of the system.

In addition, any clinical application requires the use of photon energies above 40 keV to be able to penetrate thick human body parts and a reasonable scan time. While the latter could be solved easily by using a high power source like a rotating anode x-ray tube, working with a grating interferometer at high x-ray energies is not trivial. The constraining factor preventing conventional DPC-CT systems from operating at high energies is limitations in the absorption grating fabrication process; the gold grating bars cannot easily be made thick enough to be opaque to x-rays, especially for small period gratings. This causes radiation to “leak” through the gratings and drastically reduces the visibility of the interferometer. Because of this limitation DPC-CT could not be yet made to work under optimum conditions at energies of clinical interest. To overcome this limitation Stutman *et al.*¹⁶ suggested a simple design change in the interferometer, in which the gratings are tilted along the direction of the source-detector axis, thereby presenting an effectively increased grating thickness and making it possible to operate at energies >40 keV with high visibility. Using this idea, we present the first steps to-

wards clinically compatible x-ray phase contrast CT, consisting in achieving high interferometric contrast (visibility) at a diagnostic relevant energy (~ 45 keV mean spectral value) and lowering the dose (estimated by a Monte-Carlo computer simulation) to levels approaching or similar to conventional attenuation CT. The main novelty of our work is the demonstration that it is feasible to perform DPC-CT at diagnostic x-ray energy high enough to penetrate thick human body parts. As for the dose, we were able to reach the levels comparable to conventional absorption based CT and did not exceed the clinically compatible ones by a few orders of magnitude. We achieved that by combining for the first time an interferometer having high fringe visibility at high energy with a single image per CT angle phase-scan method. This constitutes a very encouraging result by removing the main concern about the clinical applicability of x-ray DPC-CT.¹¹

2. MATERIALS AND METHODS

2.A. Experimental setup

The experiments were carried out using a symmetric interferometer setup having equal distances between the gratings, which gives the highest angular sensitivity for a given system length (see Fig. 1).¹⁷ The gratings were produced by *MicroWorks Inc*, Germany, had $5.4 \mu\text{m}$ pitch and were used in the 3rd fractional Talbot order. The nominal thickness of gold for the source grating (G0) and the analyzer grating (G2) was $100 \mu\text{m}$. The phase grating (G1) varied depending on the configuration used (more details to follow). The whole setup length was kept at approximately 1.7 m.

To compare high energy DPC-CT with low and with high fringe visibility, the setup was used in two different configurations: at a normal incidence angle [see Fig. 1(a)] and inclined at an angle $\alpha = 23.5^\circ$ [see Fig. 1(b)], creating a so-called glancing angle interferometer (GAI). Inclination of the gratings increases their effective thickness by a factor of $1/\sin(\alpha)$, thus increasing their x-ray absorption and the interferometer visibility. Due to the same effect of grating thickness variation with tilting, the nominal height of nickel structures in the phase grating should be smaller for the GAI than the normal incidence interferometer.¹⁶ In the experiment, two different phase gratings were used, producing the same π -phase shift at roughly 45 keV x-ray energy. For the GAI we used a Ni phase grating with bars of $7 \mu\text{m}$, and the normal incidence interferometer used an Au phase grating with $8.5 \mu\text{m}$ thick bars. The distances between the gratings were kept the same in both setups in order not to change the angular sensitivity.

A mini focus tungsten target x-ray tube with $60 \mu\text{m}$ spot size was used, with a current of 1 mA and voltage of 65 kVp. To produce an x-ray spectrum with mean energy of around 45 keV the tube emission was filtered by 40 mm of water, 1.5 mm of Al, and 0.125 mm of Cu. The filtering was chosen to be the same as in the soft tissue CT setup described in Sec. 2.D. As shown in Ref. 16, high interferometer visibility can be achieved at even higher x-ray energy by increasing the inclination angle of the gratings. However, we limited the glancing angle to 23.5° because the current setup suffers

FIG. 1. Schematic view of the experimental setup at a standard (a) and at a glancing angle configuration (b). The whole setup length was kept the same for both configurations. For the glancing angle configuration, the gratings were inclined at an angle $\alpha = 23.5^\circ$. It increased the effective thickness of the gratings, thus increasing their x-ray absorption and the fringe visibility. Objects and distances are not in scale.

from beam vignetting due to narrowing of the grating openings away from the center of the grating. In a practical scanner the vignetting can be eliminated using a slightly different grating design, as discussed in Ref. 18. In our experiments we chose an inclination angle, which gave a reasonable balance between visibility and vignetting. The detector consisted of a lens coupled cooled CCD detector with pixel size of $9\ \mu\text{m}$, used in a 10×10 binning mode, and placed directly behind the analyzer grating (G2). Using a high efficiency CsI scintillator screen and $f/1$ relay lens resulted in a quantum gain of more than 10 CCD electrons per x-ray, corresponding to a high DQE of ~ 0.8 .¹⁹ The effective spatial resolution at the sample position was $\sim 78\ \mu\text{m}$ (taking both magnification and source size factors into account).

2.B. Image acquisition and processing

In order to separate the absorption, refraction (phase gradient), and ultra-small scattering components in the raw CT projection images, a phase stepping method has to be used. The three components are obtained by Fourier analysis of a few interferograms acquired at slightly different analyzer grating positions. Commonly in the DPC-CT scan, the phase stepping is done separately for every rotation angle of the sample. This is inconvenient as it is time consuming from a clinical setup point of view. Instead, we used a sliding window interleaved (SWI) stepping method first demonstrated by Zanette *et al.*²⁰ with synchrotron radiation. In SWI the grating stepping is done simultaneously with rotation of the sample and the interferometric images obtained in a small angular range ($\sim 1^\circ$ – 2°) are used to obtain the absorption, phase, and scatter (dark-field contrast) information. This procedure thus makes the DPC-CT compatible with a continuous rotation of the CT gantry. All of the images shown in this paper were obtained in this way and it is the first time this approach is being presented using a conventional x-ray tube as a source. There were no special modifications made into the processing chain of interferograms in order to account for the rotating object. The SWI data are processed the same way as the standard acquisition, using the Fourier method (with eight grating “steps” or positions in our case), and resulting in the usual three types of contrast images at each angular position. The only difference

is that instead of using eight grating steps at a given CT angle, the SWI processing uses eight grating steps at successive angles to obtain one set of absorption, phase, and dark-field images. This enables doing DPC-CT with only one exposure per CT angle, with the only price paid being an effective angular averaging of the absorption, phase, and dark-field image over a small angular range (2° in our case). The whole CT scan with an angular step of 0.25° over a 200° angle resulted thus in 800 raw images and took about 2 h. For DPC-CT reconstruction we used a cone beam reconstruction code developed “in-house” and based on a Ram-Lak filter for absorption and a Hilbert filter³ for phase reconstruction. DPC-CT simultaneously reconstructs the local x-ray refractive index decrement δ (the phase contrast CT image in the following) and the local x-ray attenuation coefficient μ (the attenuation CT image). In addition, an image proportional to the ultra-small angle x-ray scattering (USAXS) coefficient can be obtained.^{1–3} The units for μ are the same as in conventional CT [$1/\text{length}$], while the refractive index decrement is dimensionless. The Hilbert filter is used because in DPC-CT the projection images represent line integrals of partial derivatives of the object’s refractive index decrement.³

2.C. Samples

To compare the image quality of a normal incidence and a glancing incidence setup we used a well-defined phantom, which mimics material composition of the structures in a human joint. The phantom follows the joint CT phantom design by Anderson *et al.*²¹ and consists of nylon, aluminum, PMMA, and water cylindrical layers placed off center, which simulate trabecular bone, cortical bone, cartilage, and joint fluid, respectively. Further on, to verify the capability of the GAI to discriminate various soft tissues and to study the influence of the exposure time/dose on soft tissue image quality we used a second phantom made of different types of fresh pig soft tissues and of a PMMA rod, placed in a plastic vial containing water or a $\sim 50\%$ – 50% alcohol/water mixture. Both phantoms were immersed in a 40 mm thick water bath during the scan to prevent strong phase jumps (“phase wrapping”) at the air-plastic vial boundary. The phantom diameters were limited to $\leq 25\ \text{mm}$ due to the limited field of view

of our present interferometer ($\sim 40 \times 40$ mm with ~ 1.6 object magnification).

2.D. Experiments

A series of various measurements was conducted to address different aspects of GAI and compare it to a standard interferometric system. First, in order to compare the standard phase stepping method with the SWI method we performed two scans of the joint phantom. The normal method used 200 CT steps of 1° and eight phase steps per each angle = 1600 projections, with 4 mAs exposure per projection, while the SWI only 800 projections with 0.25° CT step and 8 mAs exposure per projection. Each of the scans lasted approximately 2 h.

Second, to prove the strong dependence of DPC-CT image quality on visibility and the importance of having a high visibility interferometer, we performed measurements of the same phantom using the $5.4 \mu\text{m}$ grating system at normal incidence and at 23.5° glancing incidence. Different phase grating thicknesses were used to provide the same mean energy around 45 keV. This allowed obtaining very different visibility values while keeping the rest of the important interferometer parameters almost unchanged. In addition, when tilting the gratings not only the effective thickness of the grating bars increases but also the thickness of the silicon substrates. In order to keep the beam filtration the same we placed an additional $500 \mu\text{m}$ of silicon into the beam. Lastly, we kept the exposure and photon statistics at a comparable but relatively low level in both cases (several mAs per image). As before, both CT scans were performed with SWI method with total of 800 projections.

To quantify the image quality differences, the signal to noise ratios and the contrast to noise ratios were calculated as

$$\text{SNR} = \frac{|S_a|}{\sigma_a}, \quad \text{CNR} = \frac{|S_a - S_b|}{\sqrt{(\sigma_a)^2 + (\sigma_b)^2}},$$

where S_a and S_b represent measured signals, i.e., mean values in regions of 15×15 pixels chosen for each of the materials and σ_a, σ_b represent standard deviations of the mean for the respective regions. The CT reconstruction used $50 \mu\text{m}$ voxels and ten adjacent CT slices were averaged to reduce image noise. To obtain the experimental SNR and CNR values we averaged the CT images over an area of 10×10 pixels for each of the various materials.

Finally, a series of soft tissue phantom measurements at different exposure times was performed.

2.E. Monte Carlo estimation of dose

As discussed in Ref. 18 imaging of soft tissues in extremity joints is one of the possible applications for the DPC-CT technique, because for extremities the effective dose coefficients are small and because they can be more easily immobilized to allow the longer scan times expected in DPC-CT. As such, we assumed a human knee to estimate the delivered dose for the measurements with our system. We used the PCXMC 2.0

Monte-Carlo software package²² which calculates the mean values of absorbed doses, averaged over the organ volume, using a mathematical phantom or model to simulate different parts of the human body. The calculation is based on the incident air kerma and the geometry of the x-ray beam: focus-to-skin distance FSD, degrees of rotation about the body longitudinal axis and the craniocaudal direction, entry point of the beam axis, and beam height and width at this position.

To estimate the dose that would be delivered to a human knee in the conditions of our experiment, we assumed exposure of an adult knee at a FSD equal to the source-sample experimental distance of 110 cm, with a 16 cm wide by 5 cm high beam. The air kerma was computed with PCXMC assuming a 65 kVp tube voltage and 2 mm Al/0.125 mm Cu filtering, as in the experiment. The PCXMC kerma calculation is expected to be within about 30% of the correct value.²² The dose reported in our paper was finally obtained by multiplying the total experimental x-ray tube current-time product (mAs) value, by the dose per mAs computed by PCXMC with the above assumptions.

3. RESULTS AND DISCUSSIONS

3.A. Fringe visibility

Having additional optical elements in the beam, one can adjust the performance of an interferometer by optimizing its components and their parameters. As mentioned, one of the biggest influences on image quality is the fringe visibility.

Figure 2 shows the normalized intensity variation (averaged over several adjacent pixels) during the phase stepping scan for different setup configurations, from which the visibility can be calculated. Figure 2(a) shows the intensity profiles obtained with the interferometer in the $m = 3$ Talbot order, with the gratings at a normal incidence angle (90°), and inclined at 23.5° . With an analyzer grating (G2) structure thickness of $100 \mu\text{m}$ at mean x-ray energy of 45 keV we achieved mean fringe visibility of 7% and of 23%, respectively. Having the interferometer in the $m = 1$ Talbot order, using $10 \mu\text{m}$ period gratings inclined at 17° and the x-ray tube running at 55 kVp we reached a remarkably high mean visibility of 50% [see Fig. 2(b)]. To our knowledge, it is the highest visibility achieved with a conventional x-ray tube so far. Thus, using

FIG. 2. Intensity profiles averaged over an area of 10×10 pixels obtained through phase stepping scan. One can see big increase of the fringe visibility (deeper modulation) caused by the inclination of the gratings.

FIG. 3. Horizontal line profile through a visibility map (fringe visibility values calculated for every pixel). Changing duty cycle of the grating as seen by the x-ray source is a probable cause of increase of the fringe visibility away from the center of the grating.

the GAI design we were able to get substantially higher fringe visibility than up to now, making the DPC-CT technique perform close to its best.

We also observed an interesting phenomenon of visibility increasing away from the center of the grating, to nearly 60% (see Fig. 3). One possible reason is that the decreasing width of the grating openings as seen by the X-ray source increases the beam coherence. This effect was not taken into account in the simulations in Ref. 11, so the experimental visibility can be actually higher than the theoretical prediction, approaching performance comparable to that achieved with partially coherent synchrotron x-rays. This effect did not occur however for the smaller period gratings in a higher Talbot order. In addition, when using inclined gratings the transmitted intensity decreases away from the grating center because of absorption of the x-rays not going perfectly parallel to the tall and narrow grating openings but being absorbed in gold, causing lateral field of view vignetting at the grating edges.

3.B. Dose estimation by computer simulation

The PCXMC dose calculation shows that the absorbed dose for a human knee would be $1.3 \mu\text{Gy}$ per mAs in our setup. Thus for instance, a DPC-CT scan with 800 frames of 8 mAs each would deliver $\sim 8.3 \text{ mGy}$ dose to the knee. This calculation is based on a numerical model of the knee, which includes skin, muscle, bone and lymph nodes. The largest contribution to the dose comes from the radiation absorbed in the bone. For comparison the delivered dose in an absorption based cone beam CT system for musculoskeletal extremities is $\sim 6.4 \text{ mGy}$ for low-dose protocols and $\sim 15 \text{ mGy}$ for high-quality protocols.²³

The reason that our system is different from conventional absorption based CT scanners in terms of dose delivered per mAs is that the gratings absorb approximately 50% of the radiation exiting the tube and that the FSD distance is considerably larger (110 cm versus $\sim 40 \text{ cm}$ in Ref. 18 for instance). Thus, our experiments were performed with total CT exposures in the range of 1600–12 800 mAs, corresponding to a few times higher than in conventional extremity cone-beam CT, to comparable or lower than in conventional CT.

FIG. 4. Phase contrast CT axial reconstruction slices of a joint phantom, consisting of nylon (1 and 5), aluminum (2), PMMA (3), and water (4), obtained using standard (left) and SWI phase stepping method (right).

3.C. Standard phase stepping method versus SWI method comparison

Using the GAI with the gratings inclined at 23.5° and operated in the $m = 3$ Talbot order we performed measurements using the standard and the SWI acquisition schemes, at equal total exposure, as measured by the mAs product. In terms of equivalent dose delivered to the knee this would correspond to $\sim 8.3 \text{ mGy}$. Both of the images were obtained with the same inclined grating interferometer. The only difference was the use of the sliding window interlaced phase scanning method versus the standard one.

As shown in Fig. 4 and Table I, the SWI method shows higher SNR and CNR for structures close to the center of the sample. As one moves to the periphery the interlaced scan starts losing its advantage over the standard scanning procedure, because of its intrinsic tangential averaging. This can be, however, improved by further reducing the angular rotation step of the sample (not possible in our experiments due to limited angular resolution of our rotation stage). The dark spot in the center of the sample is an imperfection which occurred during extrusion of the material.

It is thus interesting to note that the strength of the SWI method does not come primarily from a dose lowering factor. Indeed, as shown in Ref. 24 there is a minimum number of photons which are needed to obtain the phase information from a periodic function. Having this in mind, SWI is preferable over the standard scanning method as it is possible to ob-

TABLE I. Signal-to-noise and contrast-to-noise ratios comparison for various materials used in the joint phantom between the standard and the sliding window interlaced (SWI) phase stepping method.

	SNR	
	Standard	SWI
Aluminum	19.61	35.43
PMMA	0.02	3.7
Water	6.14	5.73
Nylon	11.34	7.78
	CNR	
	Standard	SWI
Aluminum–PMMA	18.32	31.02
PMMA–water	4.27	6.73
Water–nylon	12.77	9.62

tain fewer images with longer exposures and to gain improved image quality in a region of interest around the center of the sample. Further on, the SWI is advantageous because, for a given total dose, it enables acquiring more photons per projection; having higher photon statistics will make in turn all the algorithms incorporated in the DPC-CT processing chain to perform better.

3.D. High versus low visibility system comparison

Figure 5 shows a comparison of phase contrast images obtained using the conventional normal incidence interferometer [Fig. 5(b)] and using the 23.5° GAI [Fig. 5(a)]. The mean visibilities were 7% and 23%, respectively. The images were obtained by averaging 15 adjacent CT slices in order to decrease image noise. While in the higher visibility case the phantom is clearly visible in the phase contrast CT, and all its layers including the PMMA, are distinguishable, the contrast between the PMMA and water layers starts diminishing in the low visibility case. Furthermore, in the lower visibility case image noise is substantially increased.

As before, to quantify the difference, the signal to noise ratios and the contrast to noise ratios were calculated [Fig. 5(c)]. Both, CNRs and SNRs are a few times higher for the high visibility system. While contrast between different materials is present in both images, the one obtained with a high visibility system has much improved quality. Since the image formation for the phase contrast system is different than for the conventional CT and the image quality depends not only on the amount of radiation illuminating the sample but also on the optical performance (aforementioned fringe visibility) of the interferometer, the presented comparison shows that only by using a high visibility interferometer one can fully take advantage of phase contrast imaging.

FIG. 5. Phase contrast CT axial reconstruction slices of a joint phantom obtained with a high (a) and low visibility system (b) and image quality comparison (c).

3.E. Soft tissue phantom at different exposure times

To investigate the noise behavior of the phase contrast CT reconstruction at different exposure times and doses we used the soft tissue phantoms described above. Contrast to noise ratio analysis and standard deviations of the signal are presented in Fig. 6. CNRs between more structures in the phantom were hard to obtain because of the low frequency noise caused by ring reconstruction artifacts (visible, e.g., in the PMMA rod). Figure 7 compares the reconstructions of a soft tissue phantom immersed in a 50%–50% water alcohol mixture and obtained at different exposure times per frame ranging from 2 to 16 mAs. The corresponding estimated dose delivered to the knee would range from ~2.08 to ~16.64 mGy.

As seen in Fig. 7, as the exposure time per frame decreases, the noise increases. For lower exposures, fine structures such as the fat filaments embedded in muscle tissue, start disappearing. However, a more detailed study using observer-dependent based image quality assessment²⁵ is necessary to determine the lowest dose at which the images would still be usable for the clinical evaluation. The image obtained at 2 s exposure time has not only increased noise but also a phase ramp artifact which is a spatially dependent offset in the shape of an inclined plane. This is because, while in the higher exposure case the peripheral areas outside the object are used to correct for any phase offsets or ramps within the frame, in the lower dose example the algorithms do not perform as well because of substantially decreased photon statistics. The

FIG. 6. CNRs and standard deviations of the signal of the phantom's structures for different exposure times.

FIG. 7. Axial slices of PC-CT (top row) and absorption-CT (bottom row) reconstruction on the soft tissue phantom in 50% ethanol–50% water mixture, at different exposure times per frame resulting in different estimated dose delivered to the sample (16 s corresponds to ~ 16.64 mGy and 2 s to ~ 2.08 mGy to a human knee).

photon statistics at the periphery of the phantom is further decreased by the grating vignetting. Nonetheless, the soft tissue contrast is much higher for DPC-CT than absorption-CT even at very low (estimated by Monte-Carlo simulation) dose (~ 2.08 mGy), while, for example, the dose during a screening mammogram is usually ~ 3 mGy.²⁶

3.F. Soft tissue phantom filled with water

In Fig. 8 we present yet another soft tissue phantom of similar composition, but this time filled with water instead of alcohol–water mixture. Having various soft tissue types in water, the phantom has refractive indices range very close to what could be expected in a CT scan of a human body.

The images were again obtained with a SWI scan with 8 mAs per frame and a total of 800 frames, taken in a 0.25° step CT scan over 200° . The phase contrast CT image in Fig. 8(a) is compared to absorption CT images processed in two different ways. In Fig. 8(b) the absorption CT image is

obtained using angular averaging over eight angular CT steps, similar to the phase image in Fig. 8(a). In Fig. 8(c), however, the absorption based image is obtained from a SWI scan as in a regular CT scan, i.e., without any angular averaging of the projections which makes for tangential blurring in the phase image. The first conclusion from Fig. 8(a) is that having a system with high visibility (23% mean) we were able to decrease the estimated dose to almost clinically compatible levels for extremities, without significant loss of phase information or of spatial resolution. There is an observable increase of noise compared for instance to the image in Fig. 7 (the 16 mAs one) but the phase information is not lost and there is still an evident superiority of the phase contrast CT image over the absorption CT image. One can easily distinguish between various tissue types and while the tendon and the PMMA rod are almost invisible in the absorption CT image, they show up nicely in the phase contrast CT image. It is important to note that the spatial resolution in our images ($50 \mu\text{m}$ voxels) is much higher than that used in conventional CT scans of

FIG. 8. Axial slices of phase (a) and absorption (b) and (c) CT reconstruction of the soft tissue phantom filled with water at 8 mAs per frame (~ 8.32 mGy to a human knee).

human tissue nowadays. This, together with the inherently low absorption contrast for soft tissues, explains the low contrast in the absorption based image.

In addition, there is a clear difference in the noise characteristics between phase contrast and absorption CT images, with the phase contrast image exhibiting predominant lower spatial frequency noise. As discussed in Ref. 28, this difference arises both from the use of different back-projection filters in the CT reconstructions (linear ramp filter for absorption data and Hilbert filter for phase data), and from the different noise transfer properties of DPC-CT compared to absorption-CT. We note also that there is much room for improvement in our CT reconstructions: to further reduce the noise in the DPC-CT images novel back-projection and noise filtering methods could be applied and recent improvements in data processing and reconstruction such as iterative or model-based reconstruction could be implemented.^{27–32}

Nonetheless, taking into account that PC-CT provides much improved soft tissue contrast while exhibiting spatial resolution considerably better than conventional CT, the results obtained in our studies at a dose almost half of that used in high-quality conventional CT (Ref. 23) show the big potential of phase contrast computed tomography.

4. CONCLUSIONS

X-ray phase contrast CT has an obvious advantage over conventional absorption CT in the form of increased soft tissue contrast. However, translating this asset into a clinical environment is not a trivial thing to achieve. One has to think about dose limits, practicability of the system and so forth. It appears as an indisputable prerequisite, however, that for clinical applications the DPC-CT interferometer must be optimized for best optical performance possible, i.e., having both high visibility and high angular sensitivity, at high x-ray energy. As discussed in Ref. 11, performing measurements with a system having low visibility and compensating for this by using a high x-ray dose will not be clinically compatible.

The results presented in our paper show the importance of fringe visibility (V) for image quality and demonstrate essential first steps towards a clinical DPC-CT system. For the first time, we present images of a biological sample taken at an estimated dose level approaching clinically acceptable limits, by optimizing the optical setup for high energy and by using sliding window interlaced scanning. Due to photon attenuation in the gratings and the use of a low current x-ray tube (max 1 mA) the scan time of our system was way beyond clinical requirements. However, assuming a high performance x-ray tube capable of continuous operation at 30 mA current, such as the Comet MXR-160HP/11 for instance, the time would be shortened to approximately 3.5 min; this value could be further decreased by using rotating anode x-ray tubes, which typically deliver from several tens to over 100 mA current. The extremity joints are thus a good place to start clinical implementation of DPC-CT because the expected longer scan times in DPC-CT can be more easily accommodated.

Further optimization of the reconstruction, processing techniques, and artifact corrections could enable PC-CT scan

for applications having more stringent dose limitations, such as breast DPC-CT. Moreover, the dose in DPC-CT could be further lowered using spectral filtering or shaping, because phase contrast imaging uses strictly only the transmitted part of the spectrum;¹⁸ the low energy radiation, which contributes strongly to the dose without contributing to the phase image, can be filtered out before entering the patient.

As major technical improvements in the grating fabrication are not expected in the near future, the glancing angle interferometer offers a simple solution for DPC-CT at x-ray energies in the clinical diagnostic range (50–120 kVp approximately), in optimal conditions of high visibility and with clinically acceptable dose. A remarkable inference from our experiments is that, with further refinements in reconstruction and denoising techniques, it appears possible to obtain enhanced soft tissue contrast simultaneously with increased spatial resolution (comparable to micro CT scanners), while still using clinically compatible dose levels. This is much more encouraging for clinical implementation of DPC-CT than the conclusions in Ref. 11, which assessed that high resolution clinical DPC-CT will not be possible. This assessment did not consider, however, the possibility of much improved optical interferometer performance. We believe that DPC-CT could indeed become a powerful complementary technique to conventional CT, to be used for specific applications which will take advantage of high resolution soft tissue imaging, such as mammography, joint cartilage imaging, and many others.

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