Experimental proof of an idea for a CT-scanner with dose reduction potential

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ABSTRACT

Preliminary results for a new CT scanning device with dose-reduction potential were presented at the SPIE Medical Imaging conference 2007. The new device acquires the Radon data after the X-ray beam is collimated through a special mask. This mask is combined with a new and efficient data collection geometry; thus the device has the potential of reducing the dose by a factor of two. In this work, we report the first complete proof of the idea using the same simplified mask of 197 detectors as last year, and a clinical C-arm with a flat panel detector to simulate the gantry. This addition enables the acquisition of two independent and complementary data sets for reconstruction. Moreover, this clinical set-up enables the acquisition of data for clinically relevant phantoms. Phantom data were acquired using both detector sets and were reconstructed with the robust algorithm OPED. The independent sinograms were matched to a single one, and from this a diagnostic image was reconstructed successfully. This image has improved resolution, as well as less noise and artifacts compared to each single independent reconstruction. The results obtained are highly promising, even though the current device acquires only 197 views. Dose comparisons can be carried out in the future with a more precise prototype, comparable to current clinical devices with respect to imaging performance.

Keywords: CT, reconstruction, scanning geometry, scanning device

1. MOTIVATION

The problem of the dose given by current CT scanners has already transcended science to become socially relevant[1,2]. There are also recent technical reports on the subject[3–5]. Indeed an increase in the dose has accompanied the substantial improvement in the quality of the images due to recent improvements of CT technology[6,7]. However, current CT devices still suffer from certain technical problems, which either limit the quality of the images or generate an unnecessary increase in patient dose. These problems are listed in the following and briefly described in Fig. 1.

- **Physical and mechanical problems**. Such as **a.** the scattered radiation and **b.** limitation in the rotation speed in order to avoid unwanted movements of the source and detector due to G-forces.
- **Geometrical problems**. Such as **c.** the non-straight form of the data interpreted as a ray and **d.** the need of scanning more than 180 degrees to aqcuire a "half scan".
- **Rebinning**[6,8]. The data acquired with the fan or cone beam are not the same as those needed by the most employed reconstruction algorithm (FBP).
- The intrinsic problems of "z-overscanning" and "overbeaming" are being currently studied by other groups[9,10].

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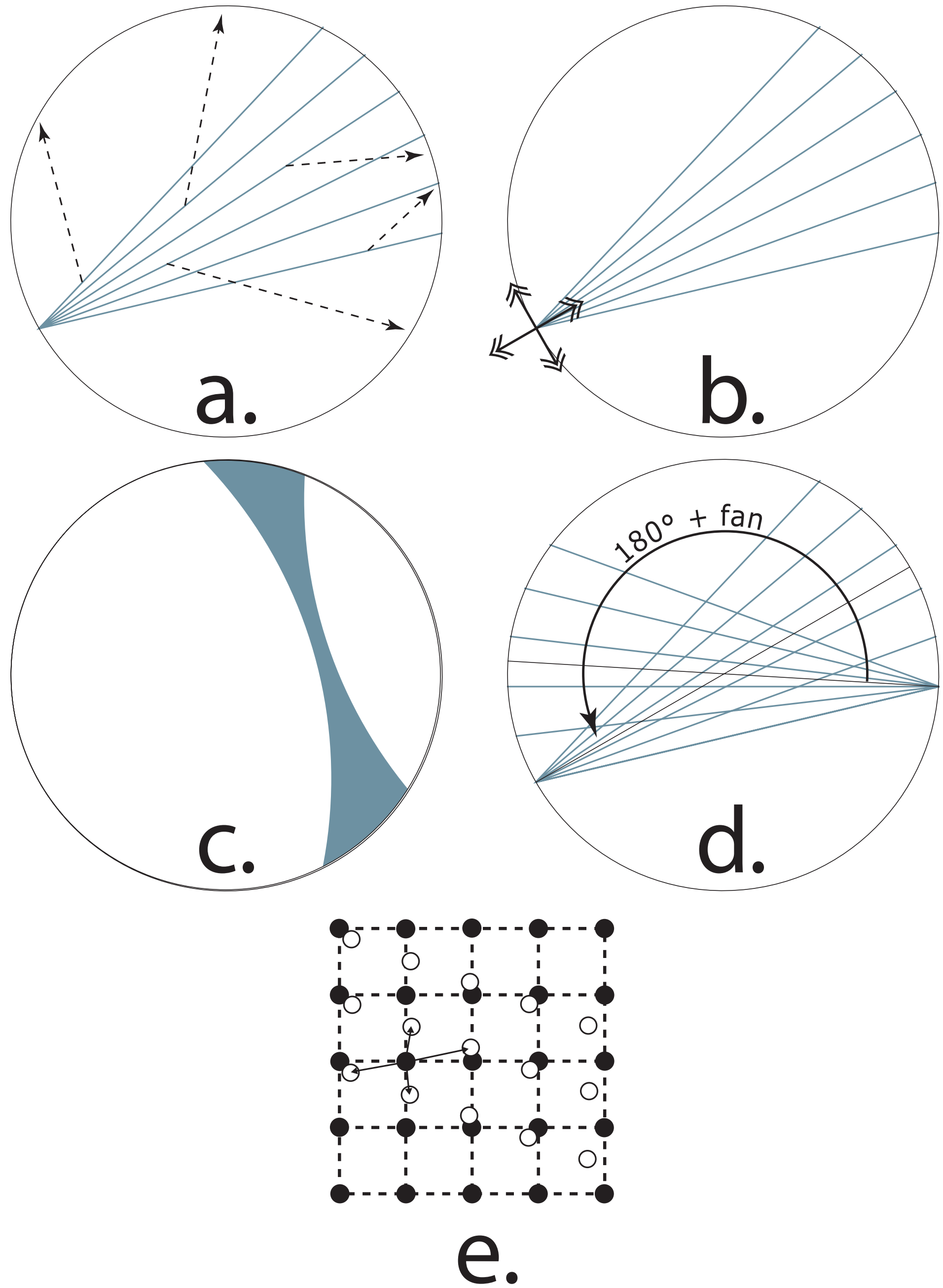

Figure 1. **Some problems of current CT devices**: **a.** Scattered radiation generates additional noise in the images. **b.** The large G-forces generated by the latest commercial models of CT devices limit the rotational speed of the gantry, thus increasing the probability of obtaining movement artifacts. **c.** The data interpreted as a straight ray by the reconstruction algorithms has a non-straight form. This shape (very exaggerated in the picture) is generated by the evolute of all trajectories travelling from the source to a detector during the time that this detector acquires a single reading. This effect limits the resolution of current devices. **d.** In order to acquire the data for a slice (a so-called "half scan"), it is necessary to irradiate the patient during a minimum gantry rotation of 180° plus the fan angle. **e.** Rebinning is the process through which the parallel data required by most reconstruction algorithms (black dots in the figure) are obtained by interpolation using the acquired fan data (white dots). This last figure is based on one found in Ref. 6.

The solution or reduction of these technical problems may lead to a more efficient use of the radiation, thus increasing the quality of the images and/or reducing the dose given to the patient. Such a solution was presented at the SPIE Medical Imaging Conference 2006[11]; It is based on a mask that can be added to any current CT device, thus solving or reducing all the problems described above without necessitating too many changes in the technology. A first prototype together with the preliminary reconstruction of a pepper were presented at the previous conference of this series[12]. However, this reconstruction used only one set of the two data sets that can be acquired with the device.

The aim of this work was to make use of a clinical C-arm to simulate the gantry, in order to be able to obtain the missing second set of data, as well as to carry out reconstructions of clinically relevant phantoms.

2. METHODS

2.1 Idea

The main idea[11] consists of collimating the X-ray beam through a circular mask of alternating detectors and windows of the same size. The detectors are shielded on the outer side of the mask. Both elements (shielding and detectors) serve to perfectly determine the ray trajectories (figure 2), as well as to protect the patient from unnecessary radiation, thus enabling **a.** direct correction for scattered radiation, since during the time that the detectors are shielded against direct radiation they receive only scattered radiation, **b.** stability against slight movements of the gantry due to G-forces, because, due to the collimation, the trajectory of a photon will always be part of a correct ray (grey stripes in figure 2), no matter where exactly the photon comes from, **c.** the acquisition of perfectly straight rays (also due to the collimation), and **d.** the scan of a slice using the dose equivalent to a rotation of 180° (since the source rotates 360° but one half of the radiation is absorbed in the shieldings). Moreover, this device acquires exactly those data that are optimally required by the reconstruction algorithm OPED[13] (Orthogonal Polynomial Expansion on Disk), a robust algorithm that does not need rebinning[14]. Thanks to the mask, the acquired fan data can be simply reordered to a sinogram of parallel data (Fig. 3).

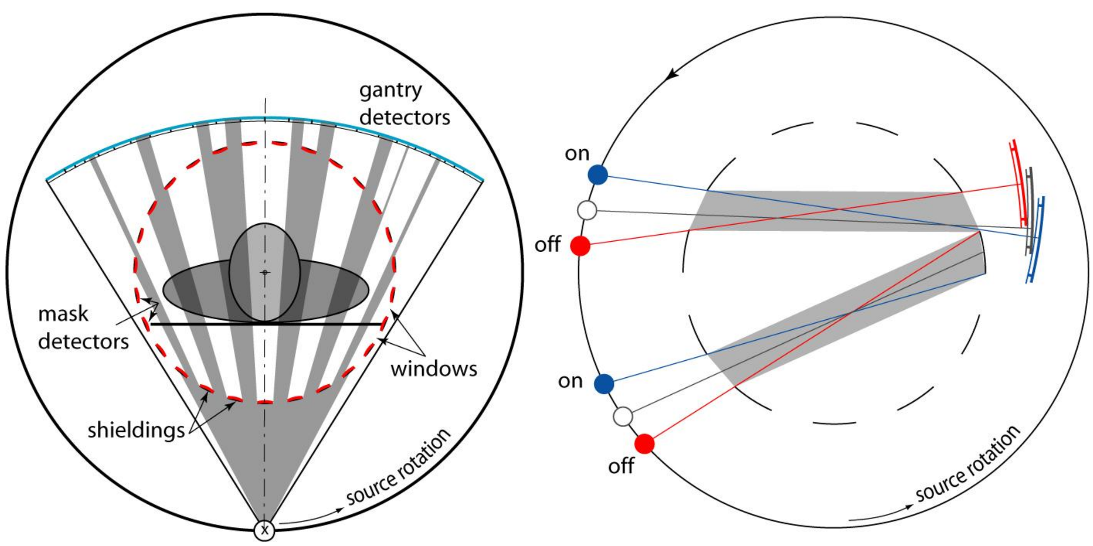

Figure 2. The idea of CT D'OR (CT with Dual Optimal Reading), presented in the Conference SPIE Medical Imaging 2006[11]. **Left**: The x-rays are collimated through the windows and are received by both mask and gantry detectors. **Right**: Every detector is recurrently exposed to radiation and shielded. One single piece of data (or "ray", represented by a grey stripe) is acquired during the time that a detector receives radiation, as explained in Fig. 1 c. for current devices. However, the shape here is necessarily straight, since the photon trajectories are collimated through the windows.

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In contemporary CT-technology, this mask may be implemented as an inner ring inside the patient tube. Such a device would then contain the mask detectors in addition to the usual "gantry" of a third generation CT, which would detect the radiation coming through the holes of the inner ring. The resulting measurements acquired by the mask-detectors and the gantry-bins (Fig. 3) allow two independent, complementary reconstructions. Hence the name CT D'OR (CT with Dual Optimal Reading).

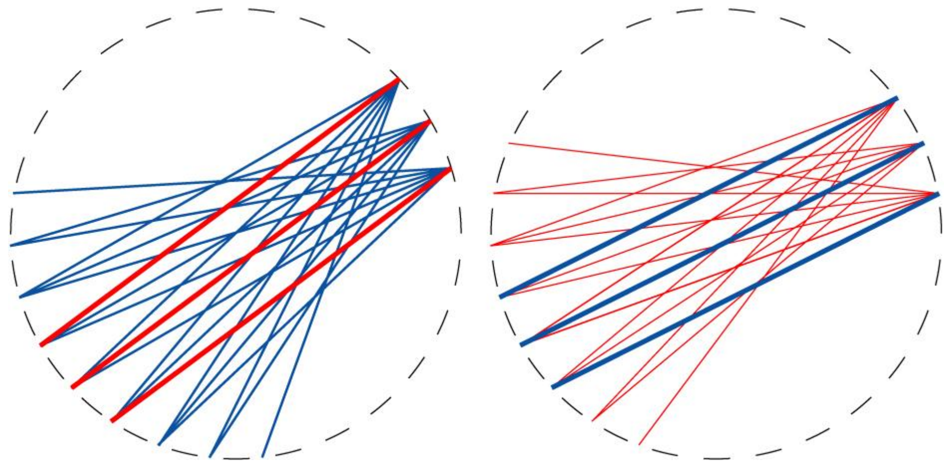

Figure 3. **Three fans of data (in the form of rays)** acquired by mask detectors (left) and by bins of the flat detector (right) after the source has rotated 40 degrees around this simplified mask of 27 detectors. In the left case, rays travel trough a window in the mask and hit a mask detector at the other side. In the second case, rays travel trough two windows of the mask and thus reach a bin of the flat detector. Note that, in both cases, every ray of a fan is parallel to one ray of the other fans (one such group of parallel rays has been marked with thick lines), which means that the parallel views (necessary for most reconstruction algorithms), are directly acquired with the scanner, thus avoiding the need for rebinning.

2.2 Prototype

We built a mask for a single-slice CT with 197 detectors. These detectors (Radcal Corp. and Scanditronix-Wellhöfer GmbH) are scintillator crystals of CsI(Tl) with an efficient surface of 1 mm^2. They are uniformly distributed on a PMMA ring of 50 cm in diameter, i.e. separated by 1.827 ($\pm$ 0.001) degrees. Every detector was shielded with 0.5 mm Pb on the outer side of the ring. From the view of the source, these shieldings are 5 mm wide, while the holes between them are 3 mm wide (Fig. 4). The whole assembly is mounted on a rotation desk (Phytron Electronics GmbH) that is controlled by a program in LabVIEW (National Instruments Corp.).

A mask-reconstruction of a pepper, carried out without the flat detector, was already presented at the previous conference of this series[12]. In order to obtain also the data for a "gantry reconstruction", we have used a clinical C-arm with a flat panel detector (ELEVA Multidiagnost, Philips GmbH). The focal spot has a size of 0.8 mm.

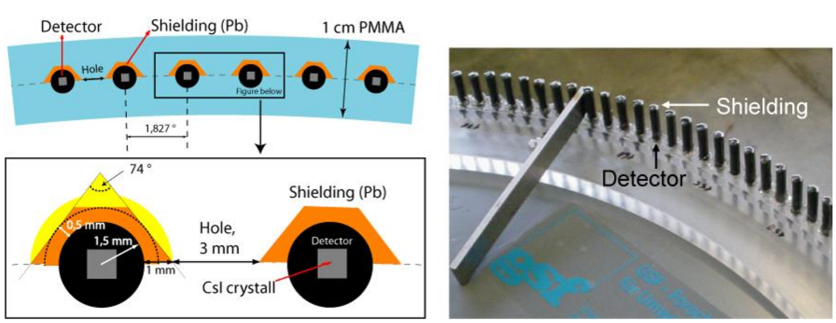

Figure 4. The design of the mask (left) and a section of the constructed mask-prototype (right).

2.3 Experiments

The mask on the rotation desk was placed between the Detector and the source of the C-arm (Fig. 5). The C-arm remained stationary and the phantom rotated 360° together with the mask at about 4° per second (actually, the mask was made to rotate 365° to ensure a full set of data over 360°). The distance between detector and source was 122 cm.

We used a photon energy quality of 96 kV, the intrinsic filtration of the device (0.1 mm Cu and 1 mm Al) and 1 mAs. Since the x-ray pulses have a duration of 4 ms, the intensity in every pulse was 250 mA. The interval between these pulses (and between acquisition of the corresponding images in the flat detector) was 33 ms. Therefore, samples of data in the ring detectors were acquired every 20 ms to exclude the possibility that two pulses were detected in the same sample. Data were acquired in sets of 30° to avoid overheating of the source.

The clinical device allows for a radiation intensity 25 times higher than the one we used for the first tests of the prototype. Therefore, it is possible now to obtain data for clinically relevant phantoms like the one shown in Fig. 6. A 3D-reconstruction of a whole head should be possible after scanning all the slices one by one.

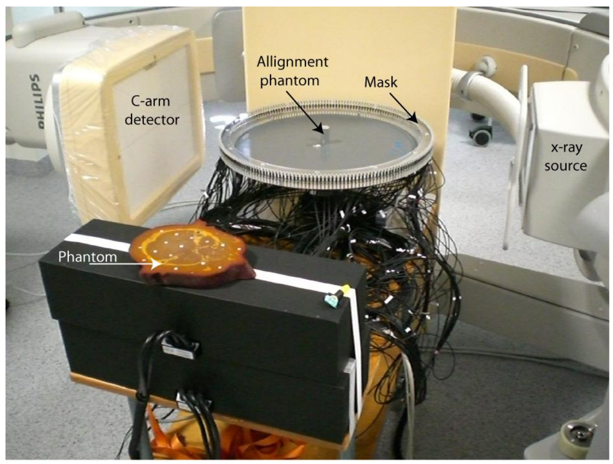

Figure 5. **The experimental set-up** during the initial alignment tests. A 2 cm wide cylinder of aluminium was placed in the middle of the mask for this purpose. The phantom finally employed in the measurements can be seen in the foreground (and also in Fig. 6). The black boxes underneath contain the photodiodes, which receive the light produced by the detectors (scintillator crystals) through optical fibres.

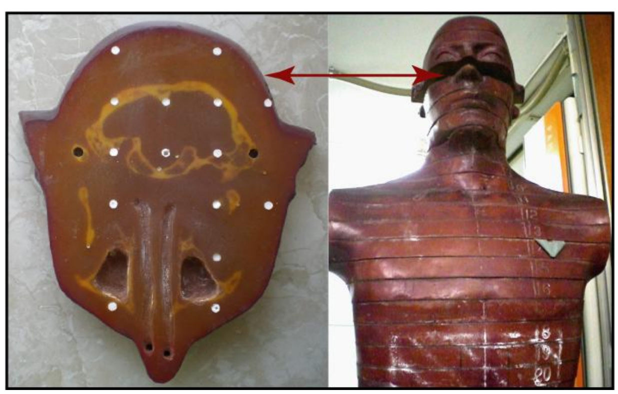

Figure 6. **The phantom** used for this study consists of a slice of a human head containing real bones and tissue-equivalent material (left) from an Alderson-Rando antropomorphic phantom (right).

2.4 Data Processing and Reconstruction

Both sets of data (mask and gantry data) can be treated separately to get independent sinograms, which can be used for independent reconstructions. However, these sinograms are complementary and have to be combined[15] to fully exploit all available data, in order to reconstruct an image that is diagnostically useful. The whole data processing can be described in the following steps:

1. **Integration.** In both sets of data, the detectors are recurrently exposed to radiation or shielded. The periods during which the detectors receive radiation are integrated into a single value, which represents a single "ray" (the data bumps in Fig. 7).

2. **Normalization.** Due to the cone beam used, the detectors that are further away of the source receive a slightly lower amount of photons than those closer to the source. A correction has been calculated and applied to the matrix after the integration.

3. **Match.** The two data sets were combined to a single set of fan data.

4. **Reorder.** The matrix of fan data does not need "rebinning", but only a simple re-ordering to the corresponding matrix of parallel data (sinogram). In the case of the gantry detectors, each ray is measured twice (once in each of two opposing directions). Therefore, the final value in the sinogram is the average of these two rays, which reduces the noise in the image. However, one half of this kind of "gantry data" cannot be acquired with the device (the trajectories of hypothetical rays flying from one detector to all other detectors), leaving empty cells in the sinogram. So far, these missing data (one fourth of the total) have been interpolated using 1-D splines*.

5. **Reconstruction**. The reconstruction has been carried out using OPED, whose detailed description can be found in references 13 and 16 .

3. RESULTS

Samples of raw data are shown in Fig. 7 for one detector of each type. The reconstructions obtained are shown in Fig. 8.

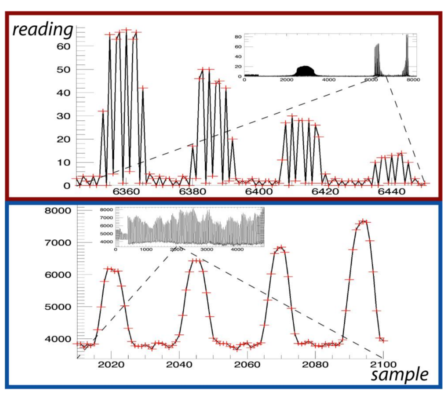

Figure 7. **One data sample of each type.** The data received by a mask detector (above) and by one bin of the "gantry" (below). A smaller region of each sample has been magnified to better visualize the bumps that represent the "rays" (grey stripes in fig. 2). These bumps are cleaner in the second case, because the data are taken synchronised with the pulses of the source. In the upper case, the higher sampling frequency generates useless null samples, which, of course, do not distort the value of the bump integrals.

*This interpolation can be avoided by using a recent algorithm developed ad hoc, which has not been implemented yet.

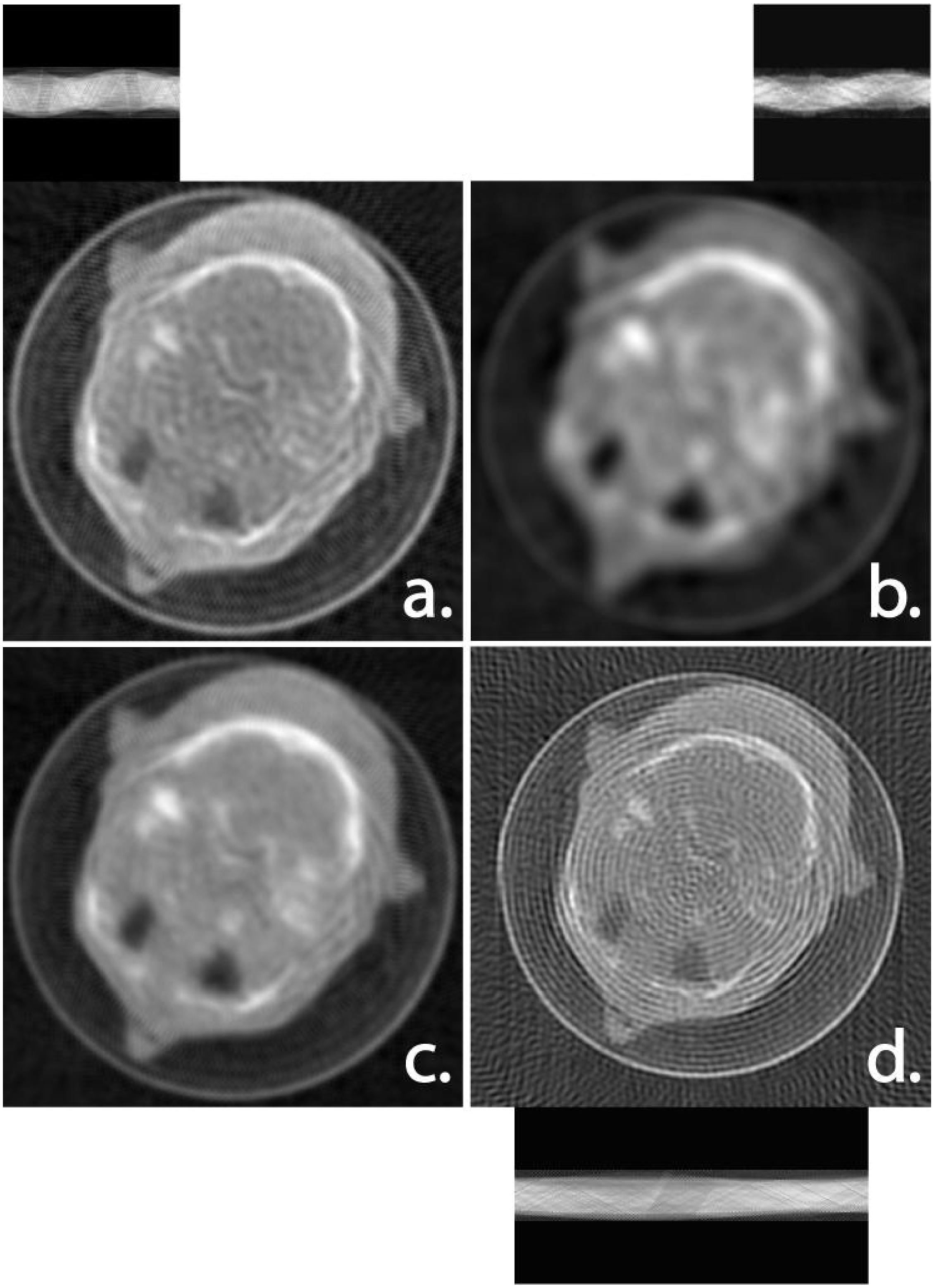

Figure 8. **The reconstructions of our head slice**. **a.** The sinogram and reconstruction from the mask detectors. **b.** The sinogram and reconstruction from the gantry detectors. **c.** The simple addition of both reconstructions (after being normalised to have an equal maximum). **d.** The combined sinogram and its final reconstruction.

4. DISCUSSION

The experimental results are satisfactory for this simplified prototype. They demonstrate that the idea of the new scanning device is certainly sound. Using 4000 mask detectors instead of only 197, the image quality achieved with a contemporary CT-scanner could be achieved. This would imply detectors and windows to have a size of approximately 1 mm, which may be a challenge for current technology, but seems to be feasible.

As expected, the first image (from the mask detectors) has a better resolution (Fig. 7 a.). The second one (from the gantry detectors, Fig. 7 b.) suffers from less noise and has better contrast (compare the dark regions of air). This is mainly due to the double collimation of the rays arriving to the gantry, which reduces the amount of scatter radiation reaching the gantry. However, the simple addition of both reconstructed images does not result in an improvement of the image quality (Fig. 7 c.). On the other hand, in the image reconstructed from the combined sinogram (Fig. 7 d.), a general quality increase in comparison to any of the other images can be observed. Nevertheless, this image suffers from clear circle-artifacts, which may be due to the different sensitivities of the two sets of detectors or to the spline-interpolation scheme. The independent sinograms have been normalized to have the same maximum value before being matched, but so far it has not been possible to achieve a continuous sinogram. WE are currently working on more careful measurements, including an empty scan, and on the implementation of the algorithm mentioned in the foot note on page 6.

A study with standard phantoms has already been planned to quantify the image quality of the reconstructions by measuring the noise and the Modulation Transfer Function (MTF) in a homogeneous and an edge phantom, respectively. In addition to this, dose measurements carried out with and without the mask, should provide an estimation of the dose saved to the patient.

However, obviously various measures have to be undertaken to further improve the quality of the reconstructed images. What our simplified prototype lacks and how we plan to improve the system in the future, is summarised below.

1. The thickness of the shields (5 mm) is larger than the size of the detectors (approx. 3 mm). As a result, 2/5 of the potential data are lost during measurements with this prototype device. Detector and shield shape will be modified in a future prototype.
2. Not all windows have exactly the same size. So, there appear circular and point artifacts in the image (even in the first image in Fig. 8). This will be avoided by performing an "empty scan" (i.e. without object), which can be used to calculate the radon data or be subtracted from the scan of the object.
3. The size of the detector's sensitive element (crystal) is about 1 mm. Despite great care, it was not possible to set them all in exactly the same plane manually (the crystals can be seen as small white dots in Fig. 9). A future prototype design will contain precise slits that enable an exact positioning of detectors and shields.

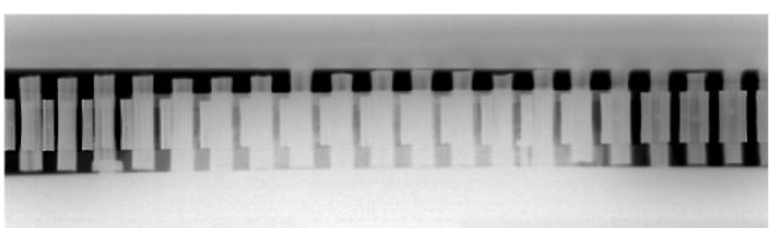

Figure 9. An example of a DICOM image acquired by the flat panel detector.

4. The position of the source focal spot is known only approximately. So, it was not possible to exactly align the position of the detector plane, with that of the focal spot. This problem, together with the previous one, increases the slice thickness of our image to about 5 mm. We will map the radiation field to find this plane more accurately.

5. The possibility has not yet been explored to analyse the scattered radiation (during the time that no direct radiation is received by each single detector) and thus potentially correct for scatter noise.
6. Measurements with standard phantoms will enable a rigorous image analysis in the future.

What we improved in comparison to the measurements of last year is listed below.

1. We obtained the second set of data using the flat panel detector of the C-arm.
2. We were able to scan the section of a head.
3. We re-glued the detectors in their holes, so that they could not move during the measurements. Apart of that, we wound the optical fibres carefully so that they do not exert any stress to the detectors.
4. Scatter radiation in the photodiodes was proven not to be a problem, but the light from the room emergency lamps. Therefore, especially designed black cases were made to keep them always in the dark.

5. CONCLUSION

The experimental results are satisfactory for this simplified prototype. They provide the full proof that the idea of the new scanning device is certainly sound. Scans of clinically relevant phantoms can be carried out and the two independent sinograms are certainly complementary. An improved device should be built and dosimetrically tested to check the potential dose reduction.

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