PET reconstruction artifact can be minimized by using sinogram correction and filtered back-projection technique

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Abstract

Filtered Back-Projection (FBP) has become an outdated image reconstruction technique in new-generation positron emission tomography (PET)/computed tomography (CT) scanners. Iterative reconstruction used in all new-generation PET scanners is a much improved reconstruction technique. Though a well-calibrated PET system can only be used for clinical imaging in few situations like ours, when compromised PET scanner with one PET module bypassed was used for PET acquisition, FBP with sinogram correction proved to be a better reconstruction technique to minimize streak artifact present in the image reconstructed by the iterative technique.

Key words: Filtered back-projection; iterative reconstruction; sinogram correction; sinogram repair; streak artifact

Introduction

New-generation positron emission tomography (PET)/computed tomography (CT) is a completely integrated system in all respects. The same system can also be used as stand alone CT scanner. CT data are required for attenuation correction of PET image to improve the quality of image.[1,2] So, CT scan is mandatory in order to generate attenuation-corrected PET images. A properly calibrated scanner can only be used for clinical studies. Even in a properly calibrated scanner, few artifacts are commonly seen on PET images due to metallic implants, respiratory motion, contrast medium, and truncation.[3-5]

Technical Note

We encountered an unusual scenario in our busy department, when the PET component of our existing PET/CT scanner (Discovery ST; GE Medical Systems, Milwaukee, Wisconsin, USA) stopped working and the service engineer temporarily started the system, but the PET component remained compromised. The cause of this was a short circuit in the preamplifier board of the ninth PET detector module; as a result, the same module was bypassed. We had three patients waiting in the post-injection area for scan, who were injected with fluorodeoxyglucose (FDG). We were in a technical dilemma whether to go ahead with the acquisition or to reschedule the injected patients. Since the referring clinician wanted to start treatment on an urgent basis, these patients were advised to undergo dedicated CT scan, as the PET scanner was not functional at the moment. However, we decided to go ahead with CT followed by PET acquisition since the patients were already injected. PET acquisition was not amounting to extra radiation exposure to the patient because the same diagnostic CT had to be used for attenuation correction and image fusion. The information from PET images was not going to be used for primary reporting, however, any additional information would be considered for reporting. We performed phantom acquisition by using image quality phantom before patient acquisition and reconstructed the image by various reconstruction algorithms available. We found Filtered Back-Projection (FBP) was able to

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produce images with optimally reduced artifact, whereas as significant streak artifact was observed in transaxial images reconstructed by iterative method [Figures 1-3]. PET images reconstructed by iterative method showed photon-deficient streak artifact in transaxial images [Figures 1A-3A, 4A-5A,4C-5C 6B-7B, 6E-7E, 8C]. The reconstruction parameter for iterative reconstruction was 22 subsets and 2 iterations and attenuation correction was performed by using Computed Tomography Attenuation Correction (CTAC) series generated by CT images. We reconstructed the image by changing the parameter to 11 subsets and 4 iterations and 22 subsets and 3 iterations, but the image quality was not changed. Same PET raw data were reconstructed using FBP and attenuation correction was performed using CTAC series generated by the CT images, which led to optimal reduction in the artifact [Figures 1B-3B, 4B-5B,4D-5D, 6C-7C,6F-7F8A]. Though CT was the primary modality for reporting, in hindsight, we realized that FBP-reconstructed PET images provided substantial information, and were used by the physician for obtaining necessary metabolic information. We imaged the same phantom after the repair of scanner and reconstructed the data by FBP and iterative reconstruction. There was mild improvement in post-repair transaxial images reconstructed by FBP [Figure 8B] and there was complete disappearance of streak artifact in post-repair transaxial images reconstructed by iterative reconstruction [Figure 8D].

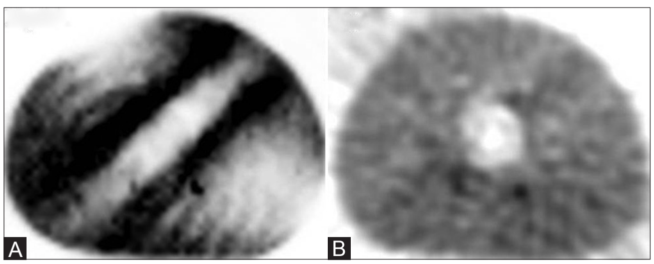

Figure 1 (A, B): Phantom images: PET transaxial image of image quality phantom showing a streak artifact (A) is reconstructed by iterative reconstruction; PET transaxial image of phantom showing optimal reduction in streak artifact (B) is reconstructed by FBP

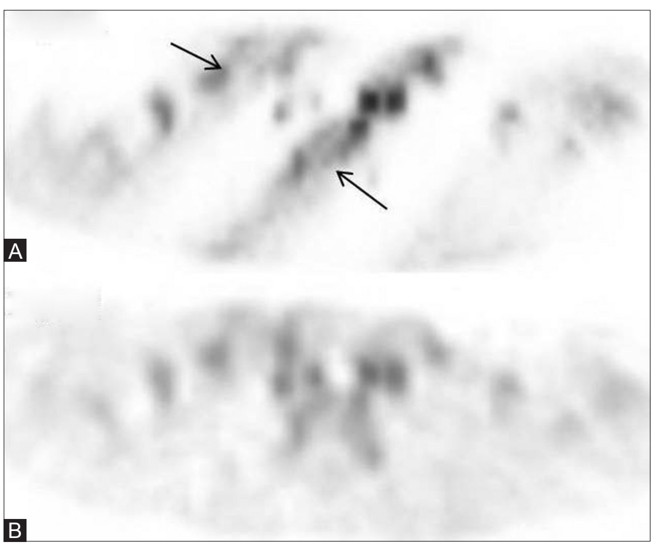

Figure 3 (A, B): Clinical images: PET transaxial image of thorax (A) is reconstructed by iterative reconstruction, showing a streak artifact (arrows); PET transaxial image of thorax (B) is reconstructed with FBP, showing optimal reduction in streak artifact

Discussion

A positron-emitting isotope is used for PET scanning. Positron and electron, by inhalation reaction, emit two anti-parallel 511 keV photons. The two photons emitted are detected on two opposite detectors in a PET scanner by coincident detection. The straight line joining these two detectors is called line of response (LOR). The true count is stored in temporary LOR binning table. Counts stored in temporary LOR binning table are binned in transverse imaging planes and transformed into sinogram by radon transformation. Image acquired by complete ring detector

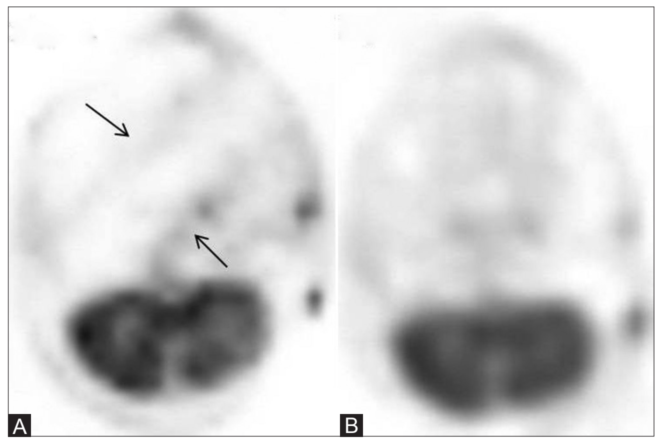

Figure 2 (A, B): Clinical images: PET transaxial image through the head (A) is reconstructed by iterative reconstruction, showing a streak artifact (arrows); PET transaxial image through the head (B) is reconstructed by FBP, showing optimal reduction in streak artifact

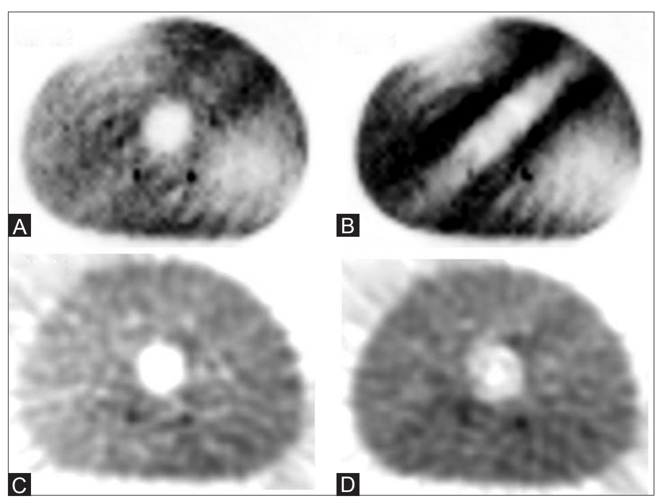

Figure 4 (A-D): Phantom images: PET transaxial image of image quality phantom showing a streak artifact (A and B) reconstructed by iterative reconstruction; PET transaxial image of phantom showing optimal reduction in streak artifact (C and D) reconstructed by FBP

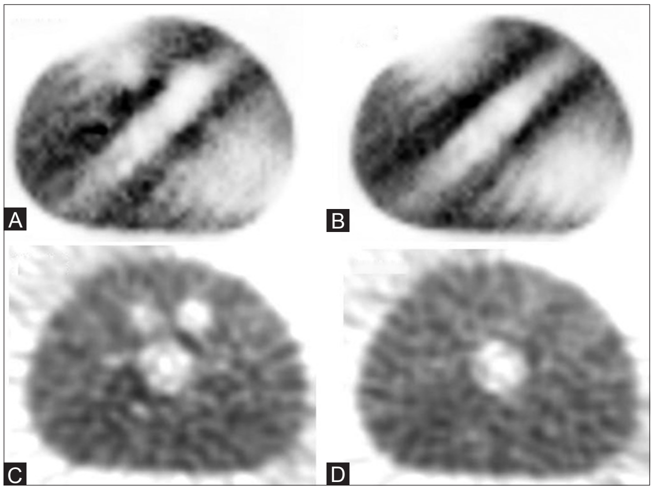

Figure 5 (A-D): Phantom images: PET transaxial image of image quality phantom showing a streak artifact (A and B) reconstructed by iterative reconstruction; PET transaxial image of image quality phantom showing optimal reduction in streak artifact (C and D) reconstructed by FBP

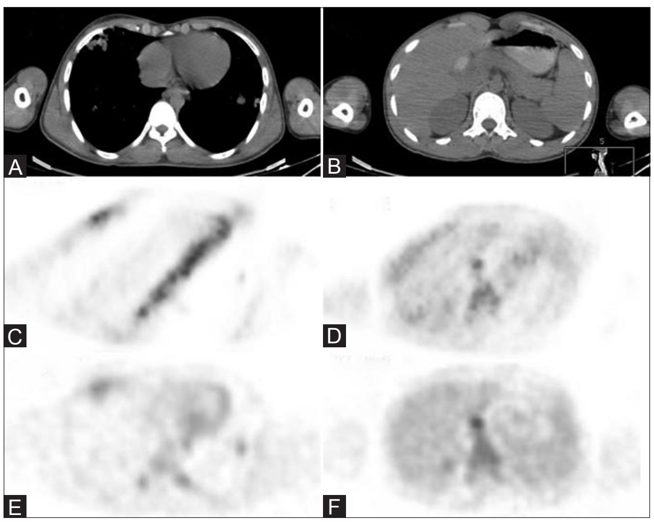

Figure 7 (A-F): Clinical images: CT transaxial images of thorax (A) and abdomen (B). PET transaxial images of thorax (C) and abdomen (D) are reconstructed with iterative reconstruction, showing a streak artifact; PET transaxial images of thorax (E) and abdomen (F) are reconstructed with FBP, showing optimal reduction in streak artifact

gives a continuous sinogram without any break. Further, the sinogram is used to reconstruct image either by FBP or iterative reconstruction technique, and various corrections are applied during reconstruction before the final images are generated. FBP is the oldest image reconstruction algorithm which has been used for years in nuclear medicine and radiology and is still used in single-photon emission computed tomography (SPECT) and CT reconstruction.[6] FBP reconstruction technique is known to introduce "noise" in the image, with star artifact in low count rate imaging like SPECT and PET, despite applying all available filters. The iterative reconstruction algorithm is an improved and more accurate reconstruction technique used in new-generation PET systems as an improved reconstruction technique.[7] PET image reconstructed by iterative method is superior and more accurate for quantification than the image reconstructed by FBP.[8] With defective detector or block of detectors, both the reconstruction techniques produce streak artifact in reconstructed image[8-10] because the LOR joining the defective detector element is not able to acquire count and assigns zero count on the LOR. When a sinogram is generated by the same acquired data, a defective sinogram is generated. The defect in the sinogram appears in the form of break in sinogram. Two sinogram repair techniques are available, which improve the quality of image by reducing the streak artifact.

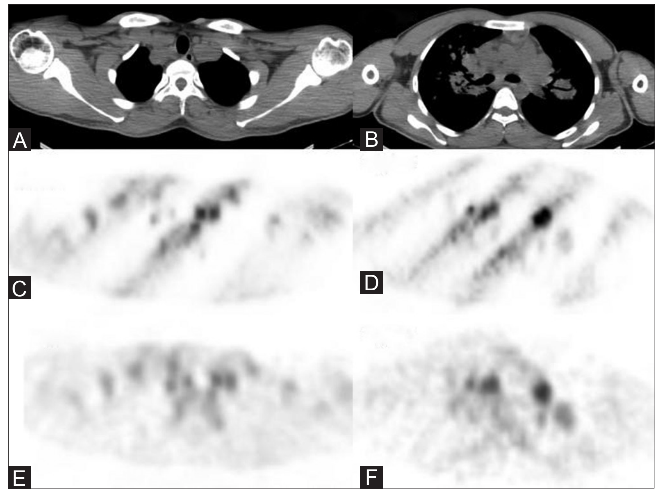

Figure 6 (A-F): Clinical images: CT transaxial image of upper thorax (A) and lower thorax (B). PET transaxial images of upper thorax (C) and lower thorax (D) are reconstructed with iterative reconstruction, showing a streak artifact; PET transaxial images of upper thorax (E) and lower thorax (F) are reconstructed with FBP, showing optimal reduction in streak artifact

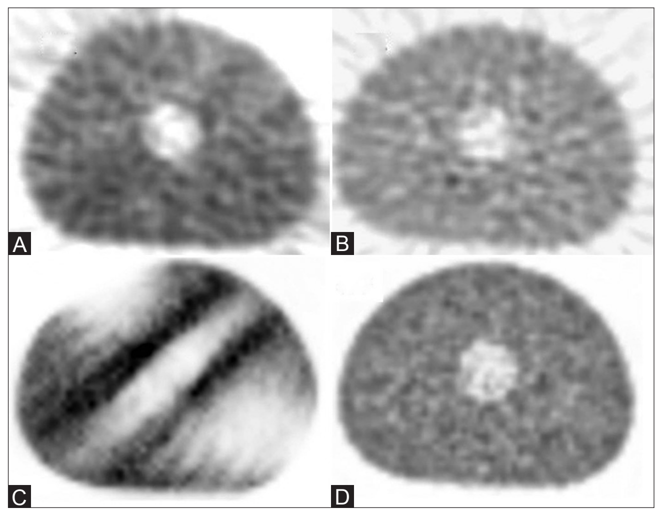

Figure 8 (A-D): Phantom images: PET transaxial image of image quality phantom showing no streak artifact (A) (defective scanner) reconstructed by FBP; PET transaxial image of phantom (B) (repaired scanner) reconstructed by FBP; PET transaxial image of image quality phantom showing streak artifact (C) (defective scanner) reconstructed by iterative reconstruction; PET transaxial image of phantom showing no streak artifact (D) (repaired scanner) reconstructed by iterative reconstruction

Sinogram repair

Identification of defective area of sinogram is mandatory. The defect on the sinogram is determined by generating a defect mask by determining thresholds for the condition frame for erroneous defect process. The values of the defective elements in the sinogram of a study are estimated mathematically and the correction method is applied to correct the sinogram. Two different methods are available for sinogram repair.

Linear interpolation method

The erroneous elements marked on sinogram in the defect mask are calculated by one-dimensional linear interpolation within columns from the values of the adjacent non-corrupted elements on the sinogram.[10] Statistical variation at these elements not marked as erroneous in the defect mask is reduced before linear interpolation by using 3 × 3 smoothing kernel K3:

$K3(i, j) = K3(i) \text{ X } K3(j)$,

$K3(i) = (0.5\ 1\ 0.5)$.

After interpolation, a second 3 X 3 smoothing using the same kernel K3 is applied at the repaired elements.

Constrained Fourier space method

The counts from sources within and outside the transaxial field of view (FOV) can be differentiated by using a matrix generated by two-dimensional Fourier transformation of a sinogram.[11] Based on this property of the two-dimensional Fourier transformation of sonogram, Karp *et al.* described the constrained Fourier space method for sinogram correction.[12] A vertical conic section on the Fourier-transformed sinogram represents the area to which counts originating outside the FOV contribute counts inside the FOV. In this context, a detector with total loss of sensitivity is a source lying outside the FOV and producing negative counts inside the FOV. The contour of the study object is estimated from the uncorrupted parts of the sinograms. The elements in the vertical conic section are set to zero and an inverse Fourier transformation of the remaining Fourier coefficients is performed in order to estimate the affected parts of the corrupted sinogram without distortion due to counts originating from the defective blocks.

But in a scenario likes ours, when compromised scanner with bypassed detector module was used for PET acquisition, a photon-deficient streak artifact, which was seen in iterative reconstructed data in transaxial image, was minimized optimally by FBP. Inherently, FBP is also known to produce the same streak artifact in the PET-reconstructed transaxial image in this scenario. FBP method of reconstruction was able to minimize this artifact in the reconstructed transaxial image and may have used one of the above-described sinogram repair techniques. Usually, the PET data sets acquired on a PET scanner with defective module produce defective sinogram with gap in the sinogram. Both the above-described sinogram correction techniques are able to correct the sinogram by filling up the gap in the sinogram. FBP is able to reduce the streak artifact in reconstructed transaxial image with corrected sinogram.

Conclusion

If an established image reconstruction technique is unable to produce optimum quality images, other available options may also be tested to circumvent the problem as described in our technical report. Knowledge of various such image reconstruction techniques is helpful in obtaining good image quality.

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