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Low-dose interpolated average CT for attenuation correction in cardiac PET/CT

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ABSTRACT

Because of the advantages in the use of high photon flux and thus the short scan times of CT imaging, the traditional ^{68}Ge scans for positron emission tomography (PET) image attenuation correction have been replaced by CT scans in the modern PET/CT technology. The combination of fast CT scan and slow PET scan often causes image misalignment between the PET and CT images due to respiration motion. Use of the average CT derived from cine CT images is reported to reduce such misalignment. However, the radiation dose to patients is higher with cine CT scans. This study introduces a method that uses breath-hold CT images and their interpolations to generate the average CT for PET image attenuation correction. Breath-hold CT sets are taken at end-inspiration and end-expiration. Deformable image registration is applied to generate a voxel-to-voxel motion matrix between the two CT sets. The motion is equally divided into 5 steps from inspiration to expiration and 5 steps from expiration to inspiration, generating a total of 8 phases of interpolated CT sets. An average CT image is generated from all the 10 phase CT images, including original inhale/exhale CT and 8 interpolated CT sets. Quantitative comparison shows that the reduction of image misalignment artifacts using the average CT from the interpolation technique for PET attenuation correction is at a similar level as that using cine average CT, while the dose to the patient from the CT scans is reduced significantly. The interpolated average CT method hence provides a low dose alternative to cine CT scans for PET attenuation correction.

1. Introduction

Positron emission tomography (PET) imaging usually includes two steps: a transmission scan for attenuation correction (AC) and an emission scan after the radiotracer is injected. The transmission scan is used as the attenuation correction for the emission scan. In recent years, PET/CT has almost replaced the traditional PET machines that use ^{68}Ge in transmission scans. The CT scan time of 100 cm scan range is less than 30 s. The photon flux of a CT scanner is typically 10^4 times that of the ^{68}Ge sources. Besides the shorter scan time, CT provides much better resolution in anatomic images. These CT advantages are paid for by higher radiation dose to the patient compared with the traditional ^{68}Ge technique. Because of the short scan time during the CT transmission scan, the patients are usually asked to hold their breath or breathe shallowly. The CT transmission images are no longer the average images over respiratory cycles. The current clinical practice performing a single helical CT scan provides only a snapshot of the respiratory cycle, whereas PET images are obtained over multiple respiratory cycles. Therefore, misalignment of the attenuation map and emission image due to respiratory motion becomes a main source of artifacts in the attenuation-corrected PET image [1,2] (Fig. 1) and can further lead to variations in PET quantification. To rectify this problem, Pan et al. [3] and others [4] reported that the use of cine average CT (CACT) can improve the accuracy of PET-CT co-registration. However, it will produce ten times radiation dose in CACT compared with clinical one shot helical CT. We therefore proposed an ultra-low dose interpolated average computer tomography (IACT) for AC on the PET cardiac data obtained from PET/CT scanners.

2. Material and methods

2.1. Image acquisition

The patients were injected with 555–740 MBq of ^{18}F-FDG and scanned for 1 h after injection. The PET acquisition was 3 min per

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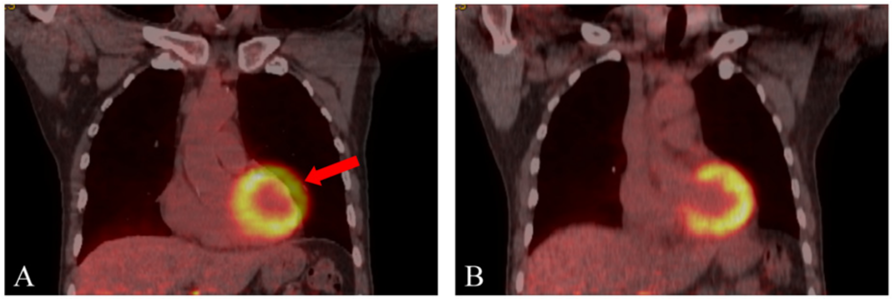

Fig. 1. Illustration of respiratory-induced misalignment between the attenuation map and emission image in CT-based attenuation for PET imaging. (A) Misregistration on PET/CT fusion image in coronal view. The arrow indicates heart borders on CT and PET emission image as unmatched, with region of misregistration corresponding to an area of artificial defect. (B) Fused PET/CT fusion image shows good co-registration associated with no defect and a normal scan.

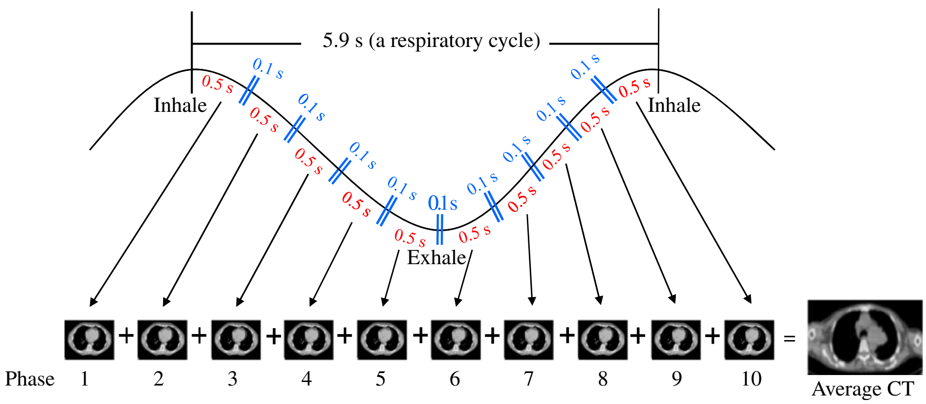

Fig. 2. Cine CT images for AC of PET emission data. Each cine scan covered 2 cm of anatomy. CACT data were acquired at 120 kV, 10 mA, 8 × 2.50 mm of X-ray collimation, 0.5 s CT gantry rotation and 5.9 s cine CT duration.

bed of 15 cm. In addition to the cine CT, two helical CT data sets were acquired at mid-inspiration and expiration phase for this study. We used an ultra-low tube current of 10 mA to minimize radiation dose for the two phases of helical CT images; the effective dose was about 1 mSv. The cine CT was taken to perform the CACT attenuation technique as a reference to the proposed IACT technique. Fig. 2 shows how the CACT technique works.

2.2. Deformable image registration

The deformable image registration algorithm used in this study was the image intensity gradient based optical flow method (OFM). This algorithm is derived from two fundamental assumptions: (1) the intensity of a point in the image does not change with time and (2) nearby points in the image move with a similar pattern [5,6].

Let a continuous and differentiable image be denoted as $f(x, y, z, t)$, where f is the gray-scale intensity at position (x, y, z) at time t. After time dt, the corresponding position shifts to $(x+dx, y+dy, z+dz)$. The function $f(x+dx, y+dy, z+dz, t+dt)$ can be expressed in a Taylor series expansion as:

$$f(x+dx,y+dy,z+dz,t+dt)=f(x,y,z,t)+\partial f/\partial x(x,y,z,t)dx + \partial f/\partial y(x,y,z,t)dy+\partial f/\partial z(x,y,z,t)dy+\partial f/\partial t(x,y,z,t)dt+\ldots \quad (1)$$

Assuming conservation of image density:

$$f(x+dx,y+dy,z+dz,t+dt)=f(x,y,z,t) \quad (2)$$

yields the optical flow equation:

$$\partial f/\partial x(x,y,z,t)vx+\partial f/\partial y(x,y,z,t)vy+\partial f/\partial z(x,y,z,t)vz+\partial f/\partial t(x,y,t)=0 \quad (3)$$

where $vx=dx/dt$, $vy=dy/dt$, $vz=dz/dt$.

Applying variational calculus a set of recursive equations is developed

$$\begin{aligned} v_x^{(n+1)} &= v_x^{(n)} - \frac{\partial f}{\partial x}\frac{\left(v_x^{(n)}\frac{\partial f}{\partial x}+v_y^{(n)}\frac{\partial f}{\partial y}+v_z^{(n)}\frac{\partial f}{\partial z}+\frac{\partial f}{\partial t}\right)}{\alpha^2+\left(\frac{\partial f}{\partial x}\right)^2+\left(\frac{\partial f}{\partial y}\right)^2+\left(\frac{\partial f}{\partial z}\right)^2} \\ v_y^{(n+1)} &= v_y^{(n)} - \frac{\partial f}{\partial y}\frac{\left(v_x^{(n)}\frac{\partial f}{\partial x}+v_y^{(n)}\frac{\partial f}{\partial y}+v_z^{(n)}\frac{\partial f}{\partial z}+\frac{\partial f}{\partial t}\right)}{\alpha^2+\left(\frac{\partial f}{\partial x}\right)^2+\left(\frac{\partial f}{\partial y}\right)^2+\left(\frac{\partial f}{\partial z}\right)^2} \\ v_z^{(n+1)} &= v_z^{(n)} - \frac{\partial f}{\partial z}\frac{\left(v_x^{(n)}\frac{\partial f}{\partial x}+v_y^{(n)}\frac{\partial f}{\partial y}+v_z^{(n)}\frac{\partial f}{\partial z}+\frac{\partial f}{\partial t}\right)}{\alpha^2+\left(\frac{\partial f}{\partial x}\right)^2+\left(\frac{\partial f}{\partial y}\right)^2+\left(\frac{\partial f}{\partial z}\right)^2}. \end{aligned} \quad (4)$$

The set of recursive equations given in Eq. (4) is applied to the inspiration and expiration CT image volumes using a multi-resolution approach to estimate the displacement field.

Originally, the optical flow method could only handle very small displacements, less than one voxel difference, limiting its applications. This problem was solved by implementing a multi-resolution technique, with larger voxels at lower resolution. The registration starts at a user given resolution level that is a 2 to the nth power multiple of the original resolution, and increases hierarchically until the finest resolution is achieved (Fig. 2).