Attenuation Correction of PET/MR Imaging

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KEYWORDS

• PET • MR • Attenuation correction • Atlas • UTE • ZTE

KEY POINTS

- Atlas and direct imaging based methods are 2 major categories of MR-based PET/MR attenuation correction.
- Atlas-based methods are accurate and robustness in brain PET attenuation correction; however, atlas-based approach cannot account for intersubject variations and is time consuming.
- Direct imaging-based MR attenuation correction methods are rapid and can account for variations across patients.
- Direct imaging with segmentation only approaches have large attenuation correction errors owing to discrete linear attenuation coefficient linear attenuation coefficient values.
- Direct imaging with segmentation and MR-computed tomography conversion has similar PET attenuation correction accuracy as the atlas-based MR attenuation correction.

INTRODUCTION

Simultaneous PET and MR imaging offers unprecedented opportunities to synergize the physiologic and molecular imaging capability of PET and the excellent anatomic and functional imaging capability of MR. This instrument opens up many possibilities for investigation in oncology, Alzheimer's disease, Parkinson's disease, and epilepsy, which are discussed in David S. Lalush's article, "MR-Derived Improvements in PET Imaging," in this issue. Simultaneous PET/MR imaging is emerging as a potential clinical and research tool for the development of noninvasive imaging biomarkers.

In PET imaging, an annihilation of an emitted positron with an electron produces two 511-KeV photons that move in opposite directions. These photons travel through the tissue before reaching PET detectors. The absorption and scatter caused by the photon–tissue interaction leads to photon attenuation.[1] The effect of photon attenuation on PET signal is described in the form of a monoexponential function as follows:

$$\frac{I}{I_0} = e^{-\mu L} \tag{1}$$

where I and I_0 are the nonattenuated and attenuated PET signals, respectively, μ and L represent the linear attenuation coefficient (LAC) and thickness of a tissue. Photon attenuation depends on the spatially varying electron density and tissue thickness. Photon attenuation can result in as high as 90% signal reduction in some regions.[2] Therefore, attenuation is by far the largest correction required for quantitative PET imaging. Small errors in estimating the attenuation correction factors may lead to significant qualitative and quantitative errors in PET images (ie, bias and artifacts).[3] PET attenuation correction methods require knowledge of the spatial distribution of tissue attenuation coefficients within the PET field of

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view. This information is represented in the form of an attenuation map (or the μ map) whose intensities represent the LAC values. In addition to tissue attenuation, some other sources of attenuation are from various hardware, such as patient table and radiofrequency coils that are placed within the PET field of view. A comprehensive review regarding hardware attenuation correction methods can be found in Paulus and Quick.[4] In this review, we focus on methods to generate patient attenuation correction maps using MR imaging.

In standalone PET systems, attenuation maps are usually estimated from a transmission scan. An external long half-life radionuclide source such as $^{68}Ge/^{68}Ga$ that emits gamma photons at a similar energy level (511 KeV) is used to acquire a transmission scan. The attenuation maps can be estimated by dividing the reference scan (blank scan, I_0) with the transmission scans. Because the number of photons emitted by the external radionuclide source is relatively low, a considerable acquisition time (approximately 10–45 minutes) is needed just for an attenuation map with an adequate signal-to-noise ratio.

Since the introduction of the first commercial PET/computed tomography (CT) scanner in early 2001, combined PET/CT replaced PET-only scanners at a rapid pace.[5] In combined PET/CT systems, a CT scan is used to provide the PET attenuation correction information.[6] The attenuation of x-rays transmitted through a patient is the source of CT contrast, and also directly related to electron density. Because the higher energy (511 keV) gamma photons in PET have a lower probability of being attenuated than the lower energy x-ray photons (80–140 keV) in CT, a piecewise linear transformation has been used to transform the CT Hounsfield unit (HU) to PET LAC values.[7,8] Compared with the PET transmission scans, CT images have higher signal-to-noise ratio and can be acquired much faster. However, it has been reported that CT-based attenuation correction led to PET quantification errors in bones.[9,10] Nevertheless, CT-based PET attenuation correction has been widely accepted as the clinical standard.

Unlike PET/CT, MR-based attenuation correction (MRAC) in simultaneous PET/MR is very challenging. MR imaging provides information on proton density and MR relaxation rates. It does not provide direct information on electron density needed by PET attenuation correction. While PET/MR imaging is FDA approved for clinical use, MR imaging-based attenuation correction methods have not been well-accepted for clinical trials. Bone has a near-zero signal in conventional MR images owing to low spin density and a rapid T2 relaxation rate, and it causes the most photon attenuation per unit volume. In contrast, air space appears similarly as the bone in conventional MR images, although it does not cause photon attenuation. Therefore, the most difficult tasks of MR-based attenuation correction are to separate bone from other tissue and air, and assign correct LACs accordingly. It has been demonstrated that improperly accounting for bone leads to a large underestimation of PET signal, particularly in tissue near bone.[11,12]

In the past several years, numerous approaches have been proposed to develop attenuation correction for PET/MR imaging. There is 1 class of method that relies primarily on PET emission data to directly estimate attenuation information through iterative joint estimation based on maximum likelihood.[13] This class of method is dubbed as maximum likelihood reconstruction of attenuation and activity. More recently, PET time-of-flight information has been incorporated into the maximum likelihood reconstruction of attenuation and activity method to improve PET attenuation correction.[14–16] MR imaging is not essential in the maximum likelihood reconstruction of attenuation and activity approaches. Regarding MR-based PET attenuation correction, there are 2 major categories of methods to generate CT like images for PET/MR imaging attenuation correction. The first category consists of an atlas-based approach.[17–20] This typically relies on a precompiled atlas of paired MR and CT images and an algorithm to generate a pseudo-CT image from patient MR images. These pseudo-CTs are converted subsequently to PET attenuation maps through the same scaling operation used in PET/CT attenuation correction. The second category of MR-based attenuation correction consists of direct MR imaging using Dixon, ultrashort echo (UTE) or zero echo time (ZTE) without using complex imaging registration and processing procedures.[20–30] In the latter approach, individual patient MR images are segmented into several tissue classes. Early efforts assigned a constant attenuation value to each tissue class. More recently, advanced methods have been proposed to derive conversion factors to convert MR signal/relaxation rates to CT HU for continuous LAC.[25,28] In this review article, we focus on the MR-based PET attenuation correction methods. The advantages and disadvantages of these methods are discussed.

ATLAS-BASED APPROACHES

The atlas-based methods usually derive a computational relationship from a group of observed CT