CT-based attenuation correction of PET images

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In the past few years there has been a strong migration from dedicated positron emission tomography (PET) scanners to hybrid positron emission tomography/computed tomography (PET/CT) scanners in hospitals worldwide. Using these combined systems, high-resolution anatomical information produced by CT adds significant information to visualization of metabolic activity delivered by PET. Therefore, these integrated systems have led to more accurate information in investigations using [F-18]fluorodeoxyglucose (FDG) in particular with regard to lesion localisation and in radiotherapy treatment planning.

PET has the reputation of being a quantitative imaging tool. This is largely based on the fact that an exact correction for attenuation of the signal due to absorption of photons in the body is theoretically achievable. Assuming a constant medium of attenuation with a uniform distribution of a positron emitter, the probability for PET to detect a coincidence event (yielding a line of response) originating from a positron emission can be expressed as $\exp(-\mu \times a)$, where μ is the attenuation coefficient of the medium for 511 keV photons and "a" is the total thickness of the medium. For example, for soft tissue μ=0.10 cm^{-1} for 511 keV photons. For a medium with a total thickness of 10 cm, this would imply a coincidence detection probability of $\exp(-0.10 \times 10) = 0.37$, or 37%. In other words 63% of the coincidence events are being attenuated due to absorption. To estimate the true amount of coincidence counts in the PET image, the original PET data need to be multiplied by a factor $1/\exp(-\mu \times a) = \exp(\mu \times a)$, also called the attenuation correction factor, in the previous example 1/0.37=2.72. Although this is an over simplification, it illustrates how to achieve an attenuation-corrected PET image theoretically. In general, the true amount of coincidence counts can be obtained by multiplying the measured PET data by the attenuation correction factor for each line of response.

With PET/CT scanners a 511 keV attenuation map (consisting of attenuation coefficients for each line of response) can be created from the CT image to correct the PET emission data for photon attenuation. There are significant advantages of using CT instead of the conventional transmission scans in stand-alone PET-systems where electronically windowed rotating rod sources as germanium-68/gallium-68, or rotating point sources as cesium-137 are used. CT data will have much lower statistical noise, which propagates into PET images. Furthermore, CT data can be acquired much more quickly than a conventional PET transmission scan, which reduces the time needed for overall whole-body PET scanning by up to 50%.

However, especially due to the use of CT for attenuation correction, movement of the patient and movement of the patient's organs, the use of combined PET/CT systems is prone to artefacts and pitfalls. This is a consequence of the fact that the attenuation coefficient depends on the tissue type, and is different for CT energies (up to 140 keV) compared to the PET photon energy (511 keV). This is illustrated in figure 1. Clearly the attenuation coefficients measured with x-ray CT must be converted to the appropriate values at 511 keV. An extra difficulty in this process of conversion is that PET uses mono-energetic 511 keV annihilation photons, whereas the x-ray source in CT emits photons with a broad energy spectrum from typically 40 to 140 keV. Usually, segmentation methods are applied to separate the CT image into regions corresponding to different tissue types (for example, soft tissue, lung and bone). The CT image values, expressed in Hounsfield Units, for each tissue type are then replaced with appropriate attenuation coefficients at a photon energy of 511 keV. Once the attenuation map at the correct energy is obtained, the PET emission data can be corrected for photon attenuation.

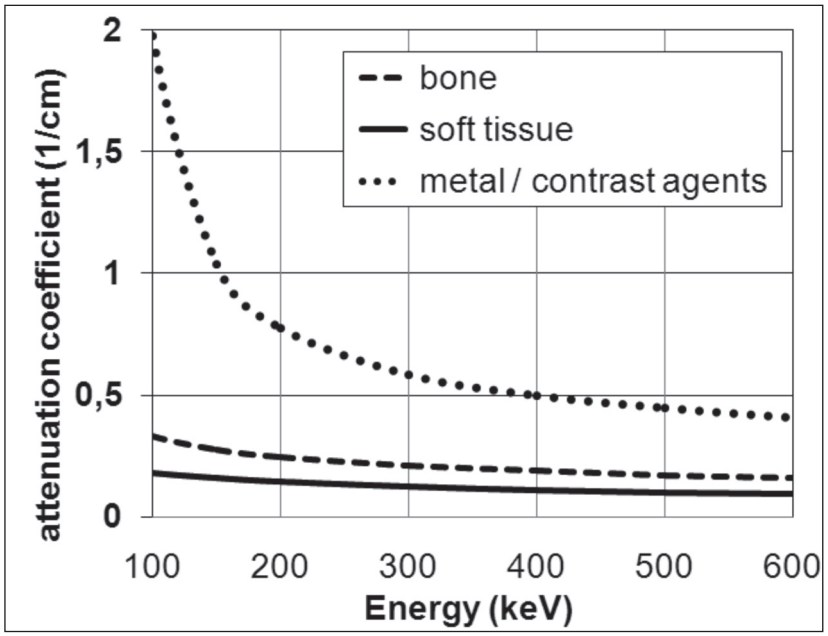

Figure 1: The attenuation coefficient as a function of the photon energy for different materials.

When the Hounsfield Units cannot be determined accurately, the attenuation map to correct the PET emission data may not be correct. Due to high photon absorption, metallic implants and contrast-enhancing agents such as iodine and barium sulphate may generate streaking artefacts on CT images, also known as beam hardening artefacts. These electron-dense areas cannot be differentiated from bone by CT image values alone. Hence, Hounsfield Units and subsequently attenuation coefficients cannot be determined unambiguously. Unlike for soft tissue or bone, the attenuation for the abovementioned electron-dense areas at CT energies is typically an order of magnitude higher than attenuation at the PET photon energy (511 keV), see also figure 1. Consequently, an overestimation of PET activity can be expected in those areas after CT-based attenuation correction, which may lead to false-positive PET findings. Therefore, for reviewing it is recommended to always use PET images that are not corrected for attenuation, as they never manifest these types of artefacts. These uncorrected PET images should be interpreted in conjunction with the corrected images.

Truncation artefacts may occur when the field of view (FOV) of the PET is larger than the FOV of the CT. These artefacts arise when positioning patients away from the "in-plane" centre of the PET/CT gantry or when imaging large patients. As a patient extends beyond the CT FOV, no attenuation values exist for the PET data in the corresponding region. This results in an underestimation of activity concentration on the attenuation-corrected PET images, potentially resulting in misinterpretation of the PET scan. The truncated regions can be recovered to a high degree when using an algorithm that artificially creates an extended FOV of the CT.

Furthermore, patient movement or respiratory movement may lead to additional artefacts in the PET images as there is a difference in acquisition time between the PET and CT. Examples are mis-registration and erroneous attenuation correction, especially in the region around the diaphragm, where a sharp transition exists on CT between liver and lung tissue. For example, an apparent contour change of the liver on CT-based attenuation-corrected PET images may be observed. In addition, there can be a reduced sensitivity for lesions in the affected area, and the PET signal will no longer be quantitative in the regions of attenuation correction artefacts, which may comprise follow-up measurements. A recommendation that can easily be implemented in clinical practice is that patients hold their breath at mid-inspiration or mid-expiration. However, maintaining an unforced expiration (exhaling) breath holding is easily underestimated. Even in an ideal situation with a fast CT scanner, accurate breath holding instructions, and an exemplary patient, the exact position of the diaphragm cannot be predicted. Upcoming methods that address motion-induced artefacts on PET are based on respiratory gating. A phased attenuation correction in respiration correlated CT/PET can lead to a more precise lesion localisation and quantification of activity.

Finally, some obvious pitfalls for integrated PET/CT systems should not be forgotten. These hybrid scanners result in unnecessary duplication of CT scans, since in most cases, a patient will already have had a diagnostic CT exam before the PET study. To avoid poor radiation safety practice, one can implement an adapted diagnostic strategy by considering, that in certain clinical conditions, obtaining a diagnostic CT before PET/CT may not be necessary. Furthermore, for many patients integrated PET/CT provides the same information that could be obtained from side-by-side imaging of a standalone CT and PET.

In conclusion, the use of integrated PET/CT systems in clinical practice has important advantages over the use of dedicated PET systems. However, important issues exist when using CT for attenuation correction. These are related to the applied attenuation coefficient which depends on the tissue type, and is different for CT energies (up to 140 keV) compared to the PET photon energy (511 keV). The wrong attenuation correction may be applied due to patient/respiratory movement, which results in misalignment of the images, the presence of metallic implants, the use of contrast enhancing agents and truncation when imaging extends beyond the CT FOV. Awareness of possible artefacts and pitfalls and being able to detect, correctly interpret and possibly solve them, is necessary for a maximum diagnostic yield.

References

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