SPECT SCATTER CORRECTION IN NON-HOMOGENEOUS MEDIA

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Abstract

Single photon emission computed tomography (SPECT) has the potential for quantitation of absolute activity concentration in vivo. The accuracy of activity estimates depends to a large extent on the accuracy of the attenuation and scatter corrections performed. This is particularly so in the thorax, where assumptions about regular object shape and constant density are inappropriate. We have developed a method of scatter correction based on convolution subtraction (CS) which takes account of variable tissue density. Rather than assuming the scatter fraction to be constant, the scatter fraction is determined at each point in the image based on measured photon transmission through the object. For comparison, a modified lower window subtraction (LWS) technique has also been developed, involving convolution of lower window data with a theoretically derived kernel, which more accurately relates lower window scatter to photopeak scatter compared with conventional LWS.

Both methods have been assessed in a phantom study using a non-uniform medium and distributed activity. The methods described were compared with the two conventional methods on which they are based. The accuracy for determining activity concentration was calculated for a 5 cm diameter "hot" cylinder, the "warm" background and a 10 cm diameter "cold" air cylinder. Quantitative accuracy of >95% was achieved in the "hot" cylinder and in the "warm" background for both the TDCS and MLWS methods, whilst all 4 methods yielded no significant reconstructed counts in the region of the "cold" air cylinder, indicating very good cold contrast.

Keywords

Quantitation; attenuation; transmission; tomography; convolution subtraction; lower window subtraction.

1. INTRODUCTION

The most commonly employed method of image reconstruction in single photon emission computed tomography (SPECT) is filtered backprojection. This method assumes that the measured projections represent line integrals of the radioactivity distribution in the object being imaged. However, due to photon attenuation and scattering within the object, the data do not conform exactly to line integrals. Without compensation for these effects, the technique is limited to qualitative rather than quantitative analysis of the reconstructed images.

The problem of correcting for attenuation and scattering is particularly difficult in the thorax where tissue density varies markedly. To date, most of the work on quantitative SPECT in the thorax has concentrated on the attenuation problem. There are a number of ways in which the accuracy of attenuation correction in non-homogeneous media may be improved. One example is to use measured attenuation coefficients (\(\mu\)) directly rather than using assumed values. This may be achieved by performing a separate CT scan (Moore, 1982) or by acquiring a transmission study using a flood source which can be acquired simultaneously with the emission scan (Bailey et al, 1987). These methods are normally used in combination with iterative or so-called 2nd order techniques (Murase et al, 1987) which generally involve refining the correction based on errors calculated from the initial, or 1st order, estimate. Using these techniques, simulation has demonstrated that quantitative accuracy of >95% can be achieved in the absence of scatter and other degrading factors (Hutton et al, 1988). However, a rigorous quantitative regime must also account for scattered photons in the correction procedure.

There are two commonly employed methods of scatter correction in SPECT. One approach is to estimate the scatter distribution by convolution of the photopeak image with a previously measured scatter
function, which usually has a mono-exponential nature. The resulting image is scaled and then subtracted from the photopeak image (Axelsson et al, 1984). The scatter function is obtained by measuring the slope of the exponential "tails" of the gamma camera response from a point source of activity placed in a scattering medium. It can be demonstrated that, in general, this method over-estimates scatter and may be improved by using the first estimate of scatter as the input to a second convolution-subtraction step to yield an improved estimate of the scatter distribution (Bailey et al, 1989). This procedure can be repeated iteratively and results in increased quantitative accuracy as well as improved contrast. These approaches assume that the scatter function (ie the exponential convolution kernel) is spatially invariant and that the scatter fraction, taken to be the ratio of scattered to total recorded counts, is constant over the entire object volume. Whilst this gives an acceptable estimate of the scatter distribution in a uniform object, recent studies using Monte Carlo simulation have shown that scatter is highly dependent on the homogeneity of the object (Ljunberg and Strand, 1990). This suggests that density values derived from transmission measurements may provide additional information which can be used to improve the estimate of scatter.

Alternatively, a second image may be recorded in a Compton scatter window (eg, 92-125keV for $^{99m}$Tc) and a fraction of this image (0.4-0.5) subtracted from the photopeak image (Jaszczak et al, 1984). This method assumes that the spatial distribution of scattered photons recorded in the lower energy window is the same as the scatter distribution in the photopeak. This is clearly not the case, since photons which are recorded in a lower energy window will have lost more energy through scattering collisions than those recorded in the photopeak. Therefore, they are likely to have undergone multiple scattering interactions or may have been scattered through greater angles giving rise to a broader scatter distribution. It has been previously demonstrated (Todd-Pokropek et al, 1984), however, that there is a relationship between the spatial distribution of scattered photons recorded in different energy windows and that the information from these data may be used to obtain an improved estimate of the photopeak scatter distribution. Lower window methods have the advantage that they inherently account for object inhomogeneity since scattered photons are recorded directly. Therefore, such techniques may be useful for comparison with other scatter correction methods, particularly in non-uniform media.

The aim of this study was to develop a scatter correction technique, based on the convolution subtraction method, which makes use of transmission measurements to improve the accuracy of the scatter estimate in non-uniform media. A modified lower window subtraction method has also been developed for comparison with the new technique.

1.2. Theory

1.2.1. Transmission Dependent Convolution Subtraction (TDSCS)

The scatter correction method proposed by Axelsson et al (1984) involves estimating the photopeak scatter image, $g_s$, by convolution of the measured image, $g_0$, with an experimentally determined mono-exponential scatter function, $s$:

$$g_s = k g_0 * s \quad (1)$$

where $k$ is the scatter fraction, assumed constant throughout the image. It is also assumed that the scatter point response, $s$, is spatially invariant, in order to satisfy stationarity. Recent simulation studies using Monte Carlo methods have shown that the space invariance of $s$ is a reasonable assumption, even in media with variable attenuation (Ivanovich and Weber, 1990). Intuitively, however, one might expect the scatter fraction to vary spatially according to regional tissue density.

The scatter fraction, $k$, at each point in the image is defined as the ratio of scattered to total recorded counts:

$$k = \frac{C_{\text{broad}} - C_{\text{narrow}}}{C_{\text{broad}}} \quad (2)$$

where $C_{\text{broad}}$ are broad beam counts (including scatter) and $C_{\text{narrow}}$ are narrow beam counts (excluding scatter). The difference between broad beam and narrow beam counts represents the scatter component.

We can also write this in terms of the buildup factor, $B$, which is defined as the ratio of broad beam to narrow beam counts at depth $d$ (Hubbell, 1963):