1. Introduction

Development of CT protocols that minimize radiation dose for specific clinical tasks is an active area of investigation, involving the tradeoff between radiation dose and image quality. The relative complex texture of CT noise, which is non-local and anisotropic (Figure 1), and the difficulty of measuring stochastic noise in the presence of anatomical structure compound this research.

Image variance mapping is a venerable tool recently that has been proposed for predicting image noise properties. This paper examines its application and limitations under clinical conditions.

2. Methods (continued)

Noise properties of CT images must be studied beginning with raw projection measurements (sinograms), defined as:

\[ A(\Theta,\phi) = \sum_{i,j} S_{ij} \delta(\Theta_{ij} - \Theta) \delta(\phi_{ij} - \phi) \]

where \( S_{ij} \) is the measured or reference flux, and the argument variables are detector, gantry, and table position. Image reconstruction is a linear process, mapping projections into images through signal processing steps such as interpolation, filtering, and back projection. These can be combined and represented as a linear operator for image formation:

\[ f(x,y) = \sum_{i,j} K_{ij} \phi(x-i,y-j) \]

where \( K_{ij} \) represents the filtered sinogram. In this paper, we focus on the variance of the logarithmic variable, which is commonly used to approximate the mean and variance of a Poisson distribution. The mean and variance of the logarithmic variable is given by:

\[ \mu = \log(\mu) \quad \text{and} \quad \sigma^2 = \log(1+z) \]

where \( z = \frac{S_{ij}}{\sigma} \)

However, this approximation relies on the fact that the measurement fluctuations must be much less than the mean measured value. In a calculation study, a similar significant errors can result. The validity of these approximations for a Poisson process are shown in Figure 2, indicating the range of signals where the approximation holds. At low signal levels, far below typical clinical protocols, the noise (and mean) will be overestimated by this approximation.

3. Results (continued)

Simulated images were created by exponentiating attenuation sinograms (including the bowtie filter and tube current modulation) and scaling to a mean flux level. This was passed through a Poisson random-noise generator, and then convolved back into attenuation. Sinograms were reconstructed using offline software.

4. Discussion

Since the measurements are essentially proportional to the mean of the signal, given that the sinograms consist of the logarithm of the measured flux, a commonly used approximation is that the variance of the logarithmic variable is:

\[ \sigma^2 = \log(1+z) \]

However, this approximation performs poorly.

5. Conclusions

The range of validity for image variance mapping was predicted and verified by simulations. For common clinical scan conditions, image variance mapping is a valuable tool for studying the effects of CT noise on image quality and developing protocols for radiation dose reduction.

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