Physical characteristics

Physical characteristics of the imaging system were modeled as follows. A simplified mono-energetic x-ray source was used. The photon energy level was set to 22 keV, this is equivalent to the effective x-ray energy of a 34 kVp Rhodium (Rh)/Silver (Ag) target/filter spectrum used for imaging a 50 mm compressed breast. An ideal point source focal-spot was used. In order to simulate the detector unsharpness, a modulation transfer function empirically measured from the reference system was used by the simulator. X-ray scatter from the test object was not considered. Only Poisson x-ray noise was added and the electronic noise was not modeled.

A calibration was performed to match the signal-to-noise ratio (SNR) in simulated images with the SNR obtained in experimentally acquired images using the GE Senographe Pristina system with automatic optimization of parameters (AOP). The calibration was repeated for FFDM and DBT acquisition modes. The adjustment method is described as follows. On the GE Senographe Pristina system, FFDM and DBT central raw projection images were acquired from two 50 mm thick phantoms (CIRS, Virginia, USA) with 100% adipose and 100% fibroglandular equivalent compositions. The phantoms were positioned side-by-side on the breast positioning table. Images were acquired at 34 kVp using a Rh/Ag target/filter combination and using AOP mode. SNR values were measured in the raw projection images (processed only with the manufacturer's gain, offset, and defective pixel correction), in 2 cm × 2 cm square ROIs at 4 cm from the chest wall. SNR was defined as $\frac{\langle \mathrm{SI} \rangle}{\sigma_{\mathrm{SI}}}$, where $\langle \mathrm{SI} \rangle$ is the average detected signal intensity per pixel and σ_{SI} is the standard deviation in the signal intensity. The experiment was then simulated. The mAs value in the simulation was adjusted so as SNR values in the simulated and experimentally obtained images were found similar. Due to restrictions of the x-ray simulator, it was not possible to exactly match the SNR for both 100% adipose and 100% fibroglandular regions. The mAs in the simulations was therefore adjusted till SNR differed by maximum 10% in both breast-tissue equivalent.

Embedding of μcalcs in FFDM and DBT raw images

The voxel-based phantoms and surface mesh-based μcalcs use different projector implementations in our x-ray simulator. One way to obtain images with both background and μcalc in a single acquisition is to voxelize the surface mesh μcalc, and insert it directly into the voxel-based phantom. This will create a single voxel-based phantom containing both the background and the μcalc. However in doing so, there might be a loss of morphological accuracy of the μcalc during the voxelization process. The accuracy loss depends on the resolution of the voxelization. At very high resolution the loss might be insignificant; but the voxel-based phantoms must also be over-sampled at the same resolution. This can greatly increase the simulation complexity in terms of processing time and phantom storage space requirement.

Therefore, a hybrid simulation method was applied to embed a separately projected mesh-based μcalc into projection images of a voxel-based phantom. The method is similar to previously described methods [36] [176]. Detail of the method is described as follows:

1. Before imaging the test objects, an adjustment of the linear attenuation coefficient of the μcalc was performed. In fact, when the background phantom and the μcalc are