overlying structures and providing limited 3D information. In addition, the artifact reduction processing method appears to yield a significant improvement in images that are corrupted by metal artifacts. Artifact reduction processing provided a high-quality image compared with conventional FBP algorithm images. Artifact reduction processing using tomosynthesis is the best solution for cases in which the high-attenuation feature causing the artifacts can be accurately segmented from the projection. Future investigations will study the ability of digital linear tomosynthesis to quantify the spatial relationship between the metallic components of these devices as well as the ability of the technique to identify bony changes of diagnostic significance.

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A Software tool for the simulation of a digital x-ray imaging system

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Keywords Simulation, Digital radiology, Voxel phantom

Purpose

The purpose of this study was the development of a computer code to simulate a complete digital radiographic system, usable for training operators and for the optimization of the radiographic techniques.

Material and methods

The used algorithm was based on a ray-tracing technique and on the x-ray attenuation law. Virtual radiological images were produced considering a punctual source and a digital detector, with a set of rays emitted towards every pixel centre of the detector. The virtual body consisted of a 3D voxel matrix generated from computed tomography series of images. The source considered was an ordinary x-ray tube, with an emission spectrum adjustable changing the peak voltage and the product of the anodic current and the exposure time. The spectrum was divided into energy bins 5 keV wide and the attenuation of each bin was calculated separately in the implemented algorithm. The linear attenuation coefficient of each voxel was estimated from the Hounsfield Unit value, considering the energy dependence of the linear attenuation coefficients for soft tissues and for the bone tissues. In order to account the actual attenuation of each beam component, the implemented algorithm calculated the intersections of the ray with the 3 dimensional voxel grid and the relative distances covered in each voxel. Finally, the photon fluence and the energy carried from the source to every pixel of the digital detector was converted in a pixel value, using the relative response function of the considered detector. The different contribution of the signal noise were also considered, adding a poissonian term related to the photon fluence distribution to other terms related to the detector response. The user interface was developed using the open source library Fast Light Toolkit (FLTK) and is shown in figure 28. The obtained image at the end of the simulation could be processed applying several look up tables in a manner similar to that of the post processing tools integrated in the software for digital detectors. In order to compare the

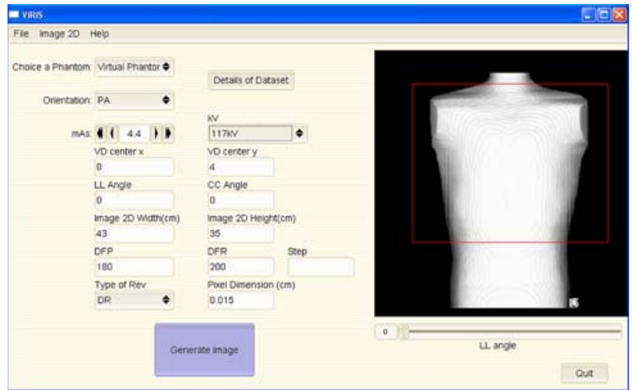

Fig. 1

simulation results with actual digital radiographies, an anthropomorphic synthetic phantom was used. Computed tomography images of this phantom were acquired with a voxel resolution of 0.7 mm x 0.7 mm x 1 mm and were used as input for the virtual body of the simulation code. Actual digital radiographies of the phantom were produced using a direct digital radiography equipment (Philips Digital Diagnost).

Results and discussion

The images of the synthetic anthropomorphic phantom produced by the simulation tool were compared with actual radiographic images of the same physical phantom. The analysis evidenced a general consistency between the simulated and the actual radiographic images, with a perfect superposition of the geometrical projection of the anatomical structures and a good correlation (r = 0.98) of the average pixel values evaluated on different areas of the image. An example of an image obtained with the simulation is shown in the attached Figure 2, and an example of an actual radiography of the anthropomorphic phantom is shown in Figure 3. The major strength of this study was the opportunity of generate radiological images keeping into account the properties of different digital detectors, such as direct radiography detectors and computed radiography

Fig. 2

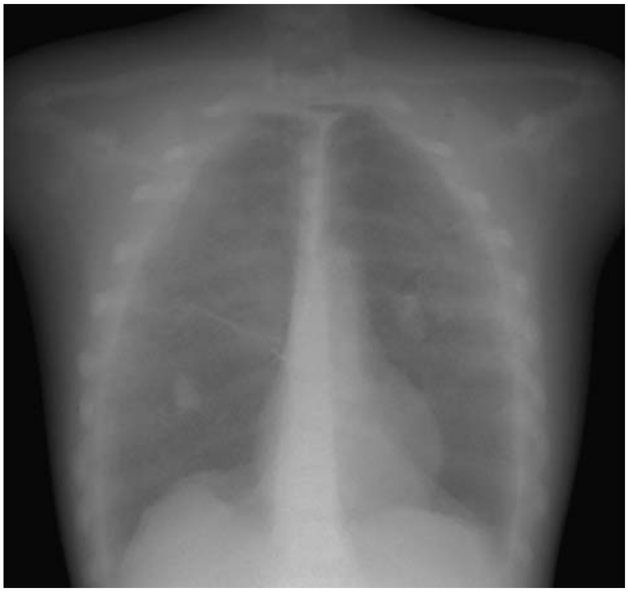

Fig. 3

detectors, considering the differences in terms of contrast and noise consequent to different choices of exposure parameters. A critical point of the tool was the time needed to produce the image, that is of the order of some decades of minutes. This time had a linear dependence with the matrix size of the generated image and it was proportional to the inverse of the square root of the phantom voxel size. As an example we could consider a single image of format 24 cm x 30 cm, with pixels of 0.15 mm x 0.15 and a voxel phantom with voxel sides of 0.8 mm. The time envolved to generate the image with a personal computer of medium performances (Notebook Dell with Windows XP, 1024 MB Ram, Intel processor 2.50 MHz, HD 60 GB 5400 rpm) was about 30 minutes. By using a pixel of size 0.5 mm x 0.5 mm the time needed to produce the image was reduced to about 3 minutes. With the continuous growing of the calculation power in the future this limitation would be expected to be less important. A limitation of the study was related to the fact that only the primary beam was considered in the produced image and no diffuse components were present. We are currently implementing the calculation of the scattering contribution to the image with a first order deterministic approach, but the time needed to produce the image is at this moment too large and an optimization of the code is necessary. However, for radiographic techniques with the use of anti scatter grids a large part of the diffusion is removed and a good similarity was observed between the simulated and the actually produced radiographies.

Conclusions

The developed software tool was found to be suitable to analyze the response characteristics of different radiological digital detectors, both for teaching purposes and for optimization studies. Further research will examine the clinical implication of this tool.

Calibration of bi-planar radiography with a minimal-size calibration object

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Keywords bi-planar radiography, geometrical calibration, 3D reconstructions

Purpose

Methods based on bi-planar radiography enable performing 3D reconstructions of bone structures (e.g. spine) with much less radiation than CT. These methods require a calibration procedure that must be executed for every examination to capture the geometrical parameters of the imaging system. In most of the published work, calibration is performed using non-standard equipments of large dimensions that enclose the patient and introduce undesirable objects into radiographs. In this paper we propose a method for geometrical calibration of bi-planar radiography that aims at minimising the impact of calibration objects on the content of radiographs, while requiring minimal adaptations to standard x-ray imaging systems.

Methods

The calibration goal is to find the values for a set of geometrical parameters that describe the imaging system at the moment that each radiograph is taken. When large calibration apparatus are not available, this is usually done by minimising the retro-projection error of a set of point matches marked in the two radiographs that compose an examination. This is accomplished using non-linear least squares minimisation, which needs an initial solution for the parameters and then iteratively updates them towards minimising the error. Unfortunately, the search space of solutions is very large, which makes difficult finding the optimum values.

In order to reduce the search space of solutions without affecting radiographs, we propose using a distance measuring device attached to the x-ray machine. This device, after calibrated, allows estimating two parameters for each radiograph. In particular, it enables to accurately calculate the distance between the x-ray source and the x-ray detector, and to have an initial estimative of the distance between the patient and the x-ray source.

In our experiments, this extension by itself only enables to determine up to scale solutions for the 3D coordinates. For correcting scale, a reference measure is needed. Such measure may be obtained by a small calibration object composed by only two radiopaque parts placed at a known distance from each other, which should be attached to the patient and be visible in both radiographs.

The proposed method was tested on 17 pairs of radiographs (size: 355.6×431.8 mm, sampling pitch: 175.0 μm/pixel), of a phantom object with dimensions of 380x380x1 mm and laser cut squares of 20.0 ± 0.1 mm evenly spaced. The corners of the squares were matched semi-automatically for each pair of radiographs. We used a laser measuring device with a typical error of $\pm$ 1.5 mm, maximum error of $\pm$ 3.0 mm, and range of operation of 0.05–50 m. The unknown calibration parameters were roughly initialised, namely the phantom was always considered to be centred on the radiograph, and its orientation was provided within a 10° resolution scale. The reconstruction scale was corrected by providing the real distance between two points of the phantom. Several combinations were tested, but always with the points separated by 40 mm. In a second experiment, uniformly distributed noise was added to every point match.

Results

Results show that when no noise is present in the point matches the errors of the reconstructed 3D coordinates of the phantom are of 0.36 mm (RMS) with 99% of these errors inferior to 0.85 mm. RMS error remains inferior to 1.0 mm when uniformly distributed noise of $\pm$ 5 pixels is added to every point match of each radiograph.

Conclusion

We conclude that the method presented here performs robustly and achieves sub-millimetric accuracy. Moreover, the requirements of the calibration object are very low when compared with other methods,