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Oversampling in the computed tomography measurements applied for bone structure studies as a method of spatial resolution improvement

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Summary

Background: Our purpose was to check the potential ability of oversampling as a method for computed tomography axial resolution improvement. The method of achieving isotropic and fine resolution, when the scanning system is characterized by anisotropic resolutions is proposed. In case of typical clinical system the axial resolution is much lower than the planar one. The idea relies on the scanning with a wide overlapping layers and subsequent resolution recovery on the level of scanning step.

Material/Methods: Simulated three-dimensional images, as well as the real microtomographic images of rat femoral bone were used in proposed solution tests. Original high resolution images were virtually scanned with a wide beam and a small step in order to simulate the real measurements. The low resolution image series were subsequently processed in order to back to the original fine one. Original, virtually scanned and recovered images resolutions were compared with the use of modulation transfer function (MTF).

Results/Conclusions: A good ability of oversampling as a method for the resolution recovery was showed. It was confirmed by comparing the resolving powers after and before resolution recovery. The MTF analysis showed resolution improvement. The resolution improvement was achieved but the image noise raised considerably, which is clearly visible on image histograms. Despite this disadvantage the proposed method can be successfully used in practice, especially in the trabecular bone studies because of high contrast between trabeculae and intertrabecular spaces.

Key words: computed tomography • image processing • image resolution • oversampling

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Background

Between different diagnostic imaging methods allowing the assessment of trabecular bone, tomographic methods like computed tomography (CT) and magnetic resonance imaging (MRI) seem to be the best choice. It is because only such methods allow three-dimensional investigation of trabeculae structure and many authors claim that structural parameters describe properly bone quality [1–6].

The three-dimensional (3D) image resolutions achieved by contemporary devices are the same order of magnitude as the geometrical parameters characterizing the trabecular bone. That is the truth if consider specially designed

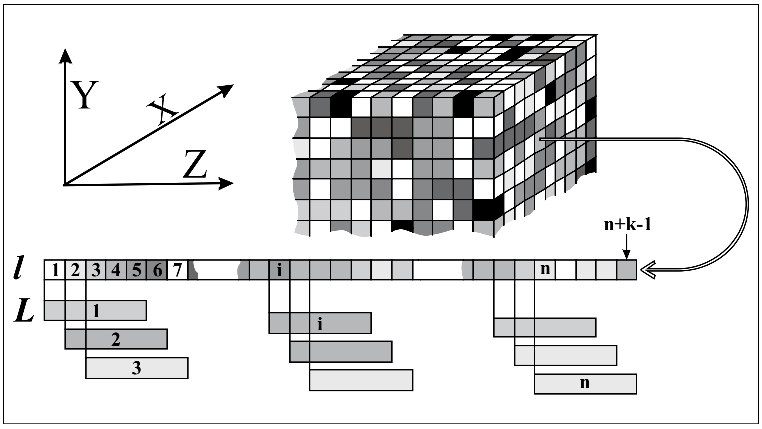

Figure 1. The proposed method idea. A space is scanned with a wide beam (**L**) and a small step. The wide voxels are the effect of averaging fine voxels (**l**). Method relies on the fine voxel values calculation using wide voxels. Only one voxel line along Z axis for particular (X,Y) position is presented.

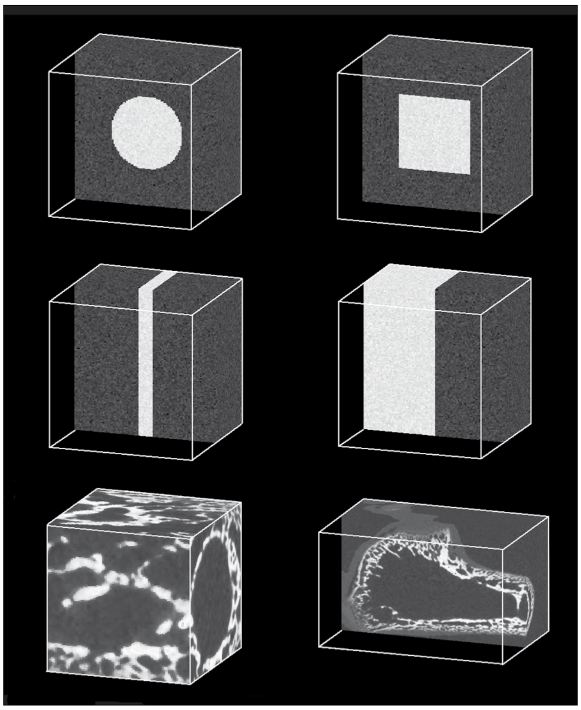

Figure 2. Few examples of the 3D images used in tests. Four simulated images of sphere, cube, "wall" and edge in the middle of the space were analyzed as well as two samples of rat bones obtained in microtomographic studies.

devices for peripheral studies (Peripheral Quantitative Computed Tomography or Quantitative Micro Magnetic Resonance Imaging) [7,8]. Some attempts are still undertaken to investigate bone structure below the spatial resolution limit [6] which is important especially in other non-peripheral locations. E.q. trabeculae structure in vertebra can not be investigated with the use of classic solution due to high signal to noise ratio [9]. In vertebral studies there is a large discrepancy between the resolution in scanning plane and in the axial direction. Usually during the image reconstruction the field of view can be defined as small enough to get the pixel size of order of about 0.1 mm. The voxel size in Z direction stays at the level of about 0.5 mm so usually the spatial resolution in Z direction is few times worse than in XY plane [5,6,10,11]. Methods allowing subvoxel structure analysis are still under interest [4,5,8].

We considered CT studies in situation when only a wide beam could be used and the resolution in XY plane is few times better than the beam width. It was shown that it is possible to get finally fine and isotropic resolution when the space is scanned with the broad beam but with small step in Z direction.

Measurements and simulations considered CT only but our conclusions can be applied in any tomographic methods characterized by resolution anisotropy.

Material and Methods

A proposed method of the CT images resolution improvement is based on two simple assumptions: (1) a CT measurement of few fine layers with a wide beam is equivalent of averaging their values in order to get the value of final larger voxel belonging to the wide layer (Figure 1) and (2) a wide beam width is an integer multiple of scanning step in the Z direction.

Because every voxel in the wide layer is the average of few fine voxels (5 in the case presented in Figure 1) a linear system of equations describes the situation from mathematical point of few for a single X, Y position along the Z direction:

$$\begin{cases} L_1 = \dfrac{1}{k}(l_1 + l_2 + \ldots + l_k) \\ L_2 = \dfrac{1}{k}(l_2 + l_3 + \ldots + l_{k+1}) \\ \ldots \\ L_i = \dfrac{1}{k}(l_i + l_{i+1} + \ldots + l_{i-1+k}) \\ \ldots \\ L_n = \dfrac{1}{k}(l_n + l_{n+1} + \ldots + l_{n-1+k}) \end{cases} \qquad (1)$$

where: $\boldsymbol{L_i}$ – the value of i-th voxel in the wide layers set along the Z axis, $\boldsymbol{l_i}$ – the value of i-th fine voxel in the fine voxels set along the Z axis, $\boldsymbol{k}$ is the number of scanning steps within one wide layer.

There are n equations in the equation system (1) but there are n-1+k unknown variables $\boldsymbol{l_i}$ so infinite number of

Table 1. The comparison between original images used in tests and the images arising as scanned with a wide beam (column A) and comparison between original images and images after resolution recovery with the proposed method (column B). As a measure of image similarity a Pearsons correlation coefficient was used.

Image	A	B
Sphere	0.943	0.973
Cube	0.962	0.985
Wall	0.846	0.981
Edge	0.970	0.993
Small bone	0.950	0.966
Big bone	0.910	0.940

solutions could be found. The solution will be single and unequivocal if k-1 parameters would be introduced. The problem is how to choose the proper parameters to get reasonable solution. In practice the problem is equivalent to the problem of guessing the values of last few fine voxels in the row. Many possibilities could be considered as a method for parameters setting but the simplest and the best, in the meaning of achieved results, is just to assume that their values are the same as the value of last wide layer voxel. The fine resolution recovery requires equation system (1) solving for every (X,Y) position in the investigated space.

For tests a set of computer simulated 3D images and 3D micro CT images of rat femoral bone taken *in vitro* were used. Some examples are shown in Figure 2.

Considering the simulated images there was software developed for the 3D artificial images generation. Image sizes were set optionally and four different images were produced: a sphere, a cube, a "wall" or an edge in the middle of the space. A Gaussian noise with adjusted variance was added to make the simulated images more realistic. Simulated images were isotropic in the meaning of spatial resolution.

Images tested in our project were subsequently processed in the second step by another peace of software aiming the simulation of CT measurements with the wide beam and with a small Z step. The CT scanning simulation was achieved with the assumption that adjacent voxels are just averaged in that process (Figure 1). The last step relied on the original resolution recovery. It was done with the use of proposed method described earlier.

The realistic images were taken in the synchrotron facility on a beamline equipped with a micro CT system (DESY, Germany, BW2 beamline) [12]. The X-ray energy was 24 keV while the achieved voxel size was about $9\times9\times9\ \mu m^3$. The real images were processed in the same way as artificial images.

In order to compare the original images and the images after resolution deterioration and subsequent recovery with the use of described method a Pearson correlation

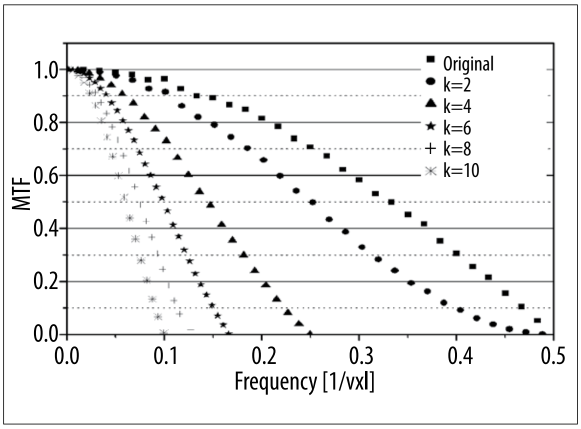

Figure 3. The MTF functions calculated for the original edge image and the images rising as scanned with a wide beam of different width.

Table 2. The frequencies for which the calculated MTF falls to 10% of maximal value. It is the measure of the real resolving power. The original edge image is consider as well as the images achieved after scanning with the beam of different widths. ***k*** defined as the ratio between the beam width and the scanning step in Z direction.

	Resolving power 1/vxl.
Original	0.47
k=2	0.40
k=4	0.22
k=6	0.15
k=8	0.11
k=10	0.09

coefficient was applied. The correlation coefficient was able to quantify images similarity but was not able to define image quality so Modulation Transfer Function (MTF) analysis [13,14] on the simulated edge images was performed. It allowed description of the real image resolution, which is correlated to the imaging method resolving power. The method of calculating MTF from the edge spread function was applied.

Results

As it was said before, the Pearson correlation coefficient was calculated to describe, if the proposed method improves the images. The correlation between the original image, image scanned with a wide beam and the image after resolution recovery was calculated. Because in real tomographic systems the XY resolution is about 5 times better than Z resolution [1,3,6,10,11] it was assumed that the scanning step in Z direction was equal to 1/5 of the beam width. The results for images shown in Figure 2 are presented in Table 1.

The MTF functions for the edge images were calculated for the original image, the image after scanning width the wide beam and the images after the resolution recovery. The k

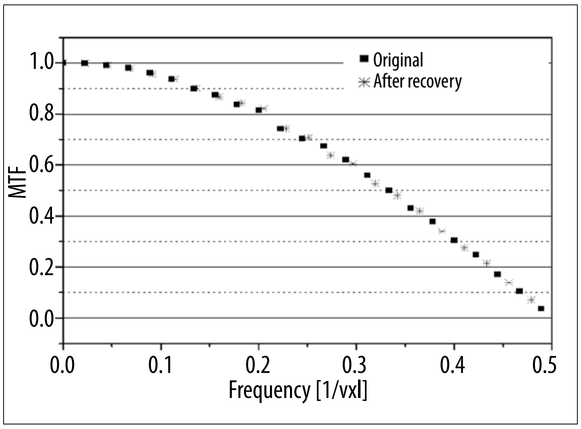

Figure 4. MTF calculated for the original edge image compared to the MTF of the image obtained after resolution deterioration with the wide beam and recovery.

blurring occurs which is typical problem in the trabecular bone analysis with the use of tomography [4–6]. Proposed method always improves the achieved images what is clearly visible if analyzing data in Table 1. The similarity measures as the correlation coefficient is always lower between the original image scanned with overlapping layer than between the original and the final one achieved by the resolution recovery procedure. The results are better for simulated images than for the realistic as one could expect.

The resolution improvement was clearly visible after the MFT analysis. The resolution, lost during scanning with wide beam (Table 2, Figure 3), was recovered almost to the same level as the image would be scanned with a fine beam (Figure 4). The MTF shape of original image and the image after resolution restoring is almost the same. Small discrepancy looks like be caused by an additional noise.

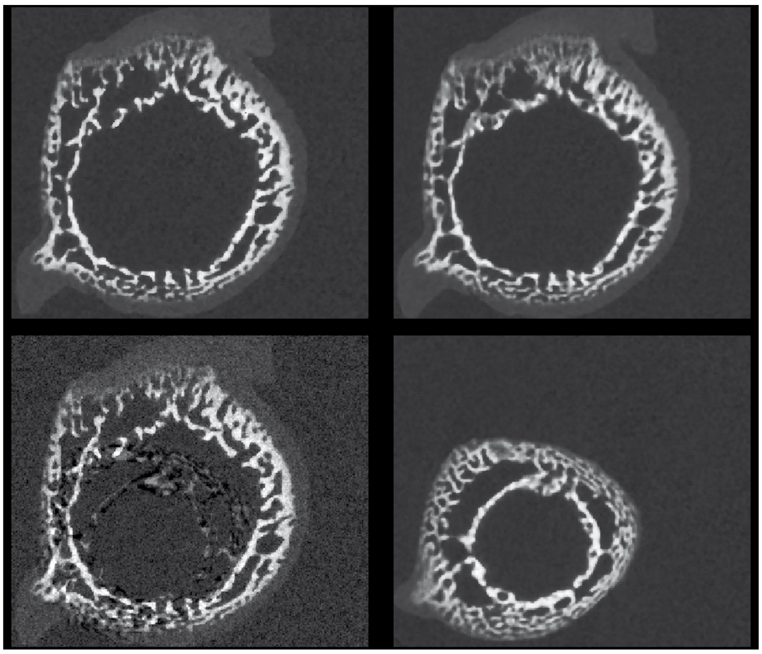

Figure 5. An example of real images taken into consideration in method tests. A rat femoral bone achieved in microtomographic measurements. The original three-dimensional image (top-left) was virtually scanned with the wide beam (the beam width was equal to five scanning steps). The image obtained (top-right) is visibly blurred comparing to the original one. It is because the low resolution layers are the average of many (five here) high resolution layers. The low resolution image series was subsequently used in order to recover the original high resolution image with the proposed procedure (bottom-left). Despite the recovered resolving power is close to the original the image noise increased. Also the shadow of the last layer of the considered scanned space is well visible (this later is shown on bottom-right).

value influence on the image resolution was investigated. The results are shown in Figure 3 and Table 2. The frequencies for which the MTF achieves 10% of maximal value are presented in Table 2. These frequencies correspond to the achieved resolving power. Because the simulated image was analyzed the voxel spacing was taken as the distance unit, so the frequencies are expressed as the reciprocal of voxel size (1/vxl).

The most interesting thing is to compare the original edge image resolution with the resolution after the Z resolution recovery. In order to do that the MTFs of both images were compared for different k values. An example for k=5 is shown in Figure 4.

Discussion

If the object is scanned with the wide overlapping beams the images became smoother but the partial volume

There is a problem of the parameters choice when the (1) equation system is solved. Few solutions of this problem were tested (e.g. the parameters can be extrapolated with the use of last few wide layers). Best results were achieved if the last few fine layers in the recovered fine voxel matrix (parameters set in fact) were just the value of the last wide layer. Such approach works well in the case of simulated images or in the case of images when the object was scanned with a margin along Z axis. The last scanned plane contains the background with noise only and the parameters choice causes only the noise rising not disturbing the image details. If there are any objects or structures cut by the 3D image border these structures shadows are propagated periodically in the recovered image as some kind of shadows (Figure 5).

An obvious effect of resolution improvement is the noise rising [11]. It is clearly visible in Figure 5 when compare the original image and the final image after resolution

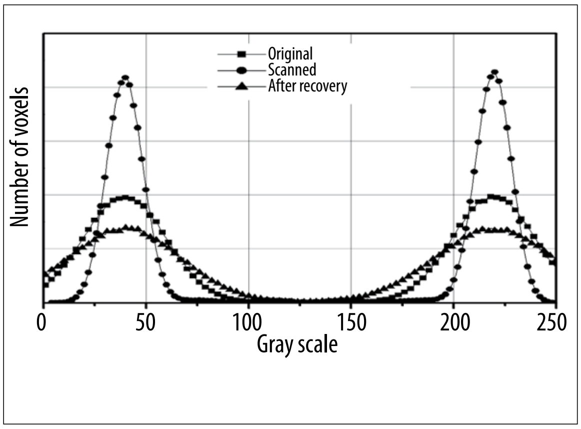

Figure 6. The comparison of image histogram of the original image, the image simulated as scanned with a wide overlapping beam and the image achieved with the proposed method. The simulated image of the edge was considered. The width of the wide beam was equal to 5 voxels.

recovery. It is also visible after the image histograms analysis. The 3D image histograms are compared in Figure 6. These are the histograms of the simulated edge image. During the simulation the Gaussian noise was added with the variance of 20. The scanning with the wide overlapping beam leads to smoother image and as a consequence the histogram maxima became narrow and high. Proposed resolution improvement method causes the maxima wider and lower than in the original image which is connected with noise rising.

The last problem is one of the disadvantages but it should be pointed that in the real measurements with fine resolution comparing to the measurements with the wide beam the noise rises also as the effect of smaller SNR caused by the decreasing voxel volume which is observed in CT as well as in MRI [6,11].

Conclusions

The proposed method of image resolution improvement is easy and works good. It can be used to equalize the image resolution in all directions when the imaging system does not ensure that. Proposed methodology requires space scanning with a small step and with wide overlapping layers. Finally achieved resolution is equal to the step size.

Because the problem of parameters choice cannot be solved unequivocally some image disturbances can be observed. These disturbances can be minimized if the object of interest is scanned with a margin in the Z direction. As the effect of presented method application the image noise is rising but it is the case of all measurements with fine resolution. The rising noise is especially not a problem for trabecular bone studies because in this case the contrast between trabeculae and intertrabecular spaces is high.

Because the method requires scanning with a large number of layers the irradiation dose in case of CT scanning rises [10] so the method would be preferably applied *in vitro* than *in vivo*.

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