To Improve the Measurement of Longitudinal Changes of Cortical Thickness and Cortical Bone Mineral Density in QCT: A Simulation Study

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Abstract

The quantification of cortical bone mineral density (BMD) and thickness in QCT images remains challenging due to the limited spatial resolution of the CT scanner. In our work, we used three different algorithms to determine changes of cortical thickness and cortical BMD and investigated their ability to detect a change of these two parameters. Another part of this study was to investigate the influence of noise on these measurements. Therefore, all simulations were performed at two different noise levels.

1 Introduction

Cortical bone is an important component of bone strength and therefore the quantification of cortical thickness and BMD at the hip, spine and forearm is of major interest in the field of osteoporosis. However, if the cortical thickness is smaller than 1 mm, the limited spatial resolution of whole body clinical CT scanners causes partial volume artifacts and as a consequence, cortical thickness may be over- and cortical BMD underestimated in quantitative computed tomography (QCT).

The consequences of spatial blurring have been studied extensively. Prevrhal et al. proposed a method based on local adaptive 50% thresholds, which is fast but leads to an overestimation of thickness for thin cortices [2]. Other methods based on thresholding also suffer from inaccuracies if cortices are thin [1]. Recent publications by Treece et al. use optimization techniques to overcome these problems [3].

The studies summarized above show that the accuracy of cortical thickness and density measurements depends on the segmentation method. Since it is essential to quantify age and drug related changes over time, we investigated the effects of three different segmentation techniques on simulated changes in cortical BMD and thickness.

2 Materials and Methods

2.1 Simulation of Image Acquisition

Segmentation of cortical bone typically is based on the BMD analysis along local profiles perpendicular to the outer bone surface. A possible method to obtain such BMD profiles from CT images is shown in Fig. 1. After an initial segmentation, which for example can be performed by volume growing, the bone surface is triangulated. For each vertex, a linear bone profile $\mathrm{BMD}(x)$ is obtained by measuring the BMD values along a line p perpendicular to the outer bone surface.

The true bone profile $\mathrm{BMD}(x)$ can be modelled as a sum of step functions of varying width and height and can be described as

$$\begin{aligned}\mathrm{BMD}(x) = \mathrm{BMD_t} + (\mathrm{BMD_c} - \mathrm{BMD_t})H(x-x_1) \\ +(\mathrm{BMD_s} - \mathrm{BMD_c})H(x-x_2)\end{aligned} \quad (1)$$

where the indices t, c, and s stand for trabecular bone, cortical bone and soft tissue. $H(x)$ is the Heaviside Function, while x_1 and x_2 determine the positions of the inner and outer bone surfaces. Therefore, the true cortical thickness is $t_c = x_2 - x_1$.

Eq. 1 is convoluted with a Gaussian function $g(x;\sigma,\mu = 0)$ approximating the point spread function of the CT scanner. The full width at half maximum (FWHM) is assumed as the scanner resolution. Therefore, the blurred profile $\mathrm{BMD_b}(x)$, simulating the density distribution within a reconstructed CT image, can be calculated as

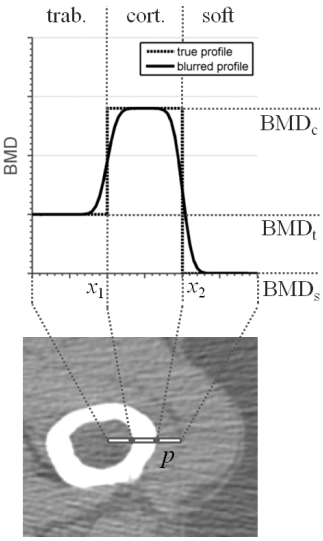


Figure 1: Profile across cortex. $\mathrm{BMD_t}$ represents trabecular, $\mathrm{BMD_c}$ cortical bone and $\mathrm{BMD_s}$ soft tissue.

$$\mathrm{BMD_b}(x) = \int_{-\infty}^{\infty} \mathrm{BMD}(t)g(x-t;\sigma)\mathrm{d}t \quad (2)$$

2.2 Estimation of Cortical Thickness and BMD

Three different algorithms are used to calculate the estimated cortical thickness te: a global threshold (GT), a local adaptive thresholds based on 50 % thresholds (LAT) and an optimization method based on Levenberg-Marquardt algorithm (OM).

GT uses global threshold values to separate soft tissue, cortical and trabecular bone. In our study we use 400 $\mathrm{mg/cm^3}$ to segment cortical bone from soft tissue and 200 $\mathrm{mg/cm^3}$ to differentiate cortical and trabecular bone. AT calculates threshold values, which are locally adjusted for each profile perpendicular to the bone surface. The positions of the outer and inner bone surfaces x_1 and x_2 are determined by calculating 50 % threshold values for each side of the cortex [2]. OM is based on a method described in [3]. The result of eq. 2 is fitted to each profile and the parameters $\mathrm{BMD_t}$, $\mathrm{BMD_s}$, x_1, x_2 and σ are determined using the Levenberg-Marquardt method. $\mathrm{BMD_c}$ is measured in the shaft below the lesser trochanter where $t_c \gg$ FWHM and therefore cortical intensity is not affected by partial volume artifacts.

To estimate cortical density $\mathrm{BMD_e}$ at a location of interest, the blurred density profile $\mathrm{BMD_b}(x)$ is integrated between the edges x_1 and x_2 and divided the result by t_e [2]:

$$\mathrm{BMD_e} = \frac{1}{t_\mathrm{e}} \int_{x_1}^{x_2} \mathrm{BMD_b}(x)\mathrm{d}x \tag{3}$$

2.3 Simulation Parameters

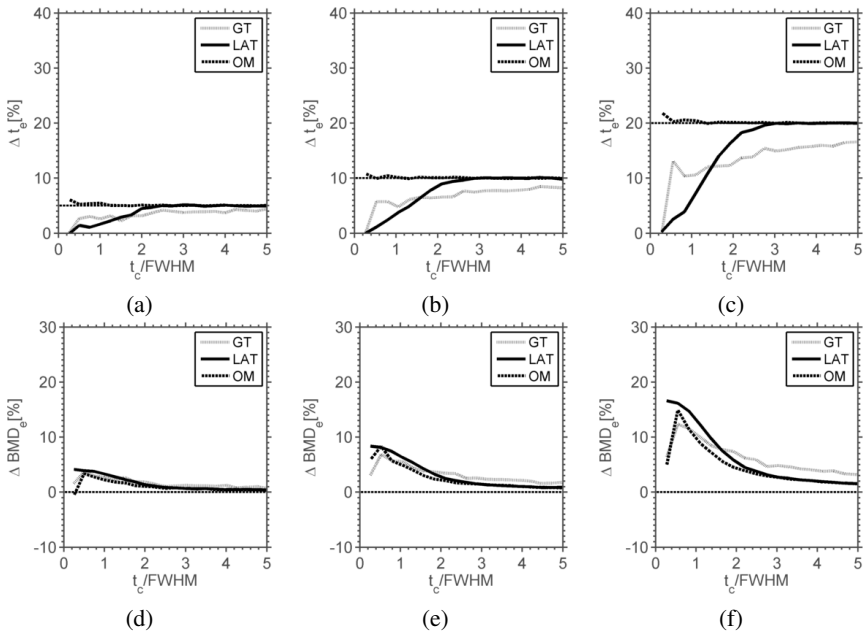
We simulated 2.5 %, 5.0 % and 7.5 % increases of $\mathrm{BMD_c}$ and 5 %, 10 % and 20 % increases of t_c, which may also occur in practice, by varying the height and width of the true bone profile $\mathrm{BMD}(x)$ for different initial cortical thickness values. Following this, $\mathrm{BMD}(x)$ is convoluted with a Gaussian distribution (see eq. 2) to create the blurred profile. Finally, the resulting curve is discretized to simulate a voxel size (s) and Gaussian noise of standard deviation σ_noise added.

Each profile was simulated 20 times, the changes Δt_e and $\Delta\mathrm{BMD_e}$ were estimated using the methods described in 2.2 and the results compared with the known true values. Baseline $\mathrm{BMD_c}$ was assumed to be 1400 $\mathrm{mg/cm^3}$. The following parameters were kept constant during the simulation process: $\mathrm{BMD_t} = 75\ \mathrm{mg/cm^3}$, $\mathrm{BMD_s} = 0\ \mathrm{mg/cm^3}$, $\mathrm{FWHM} = 0.5$ mm and $s = 0.25$ mm.

3 Results

3.1 Variation of Cortical Thickness

The effects of an assumed longitudinal 5 %, 10 % and 20 % cortical thickness increase on measured changes (Δt_e and $\Delta\mathrm{BMD_e}$) with zero noise and two different noise levels are shown in Fig. 2 as a function of $t_\mathrm{c}/\mathrm{FWHM}$.



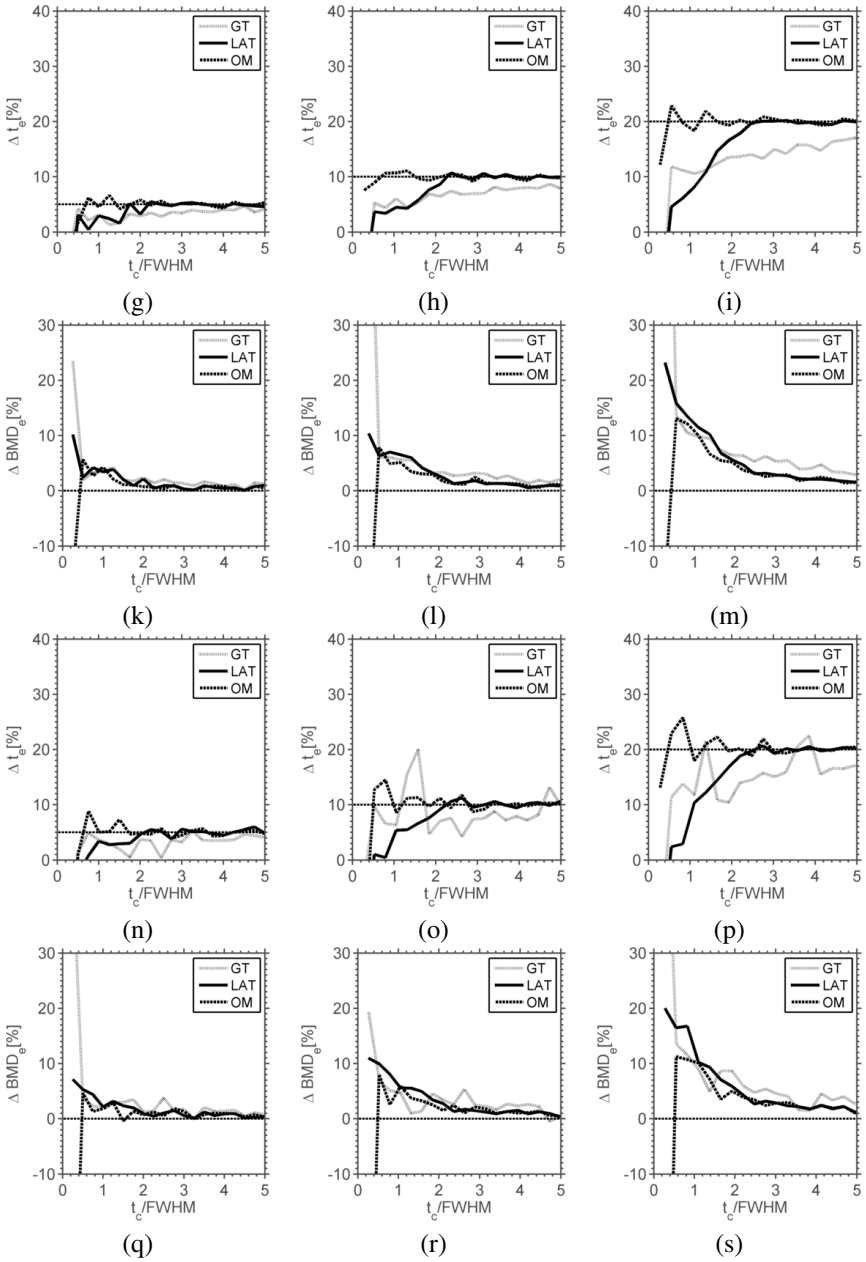


Figure 2: **Increase of t_c.** Mean values of estimated changes of Δt_e ((a) to (c), (g) to (i) and (n) to (p)) and ΔBMD_e ((d) to (f), (k) to (m) and (q) to (s)) as a function of $t_c/FWHM$ for an assumed 5 % (first column), 10 % (second column) and 20 % (third column) increase of true cortical t_c. (a) to (f) were simulated for $\sigma_{noise} = 0\ mg/cm^3$, (g) to (m) for $\sigma_{noise} = 30\ mg/cm^3$ and (n) to (s) show the results for $\sigma_{noise} = 37\ mg/cm^3$

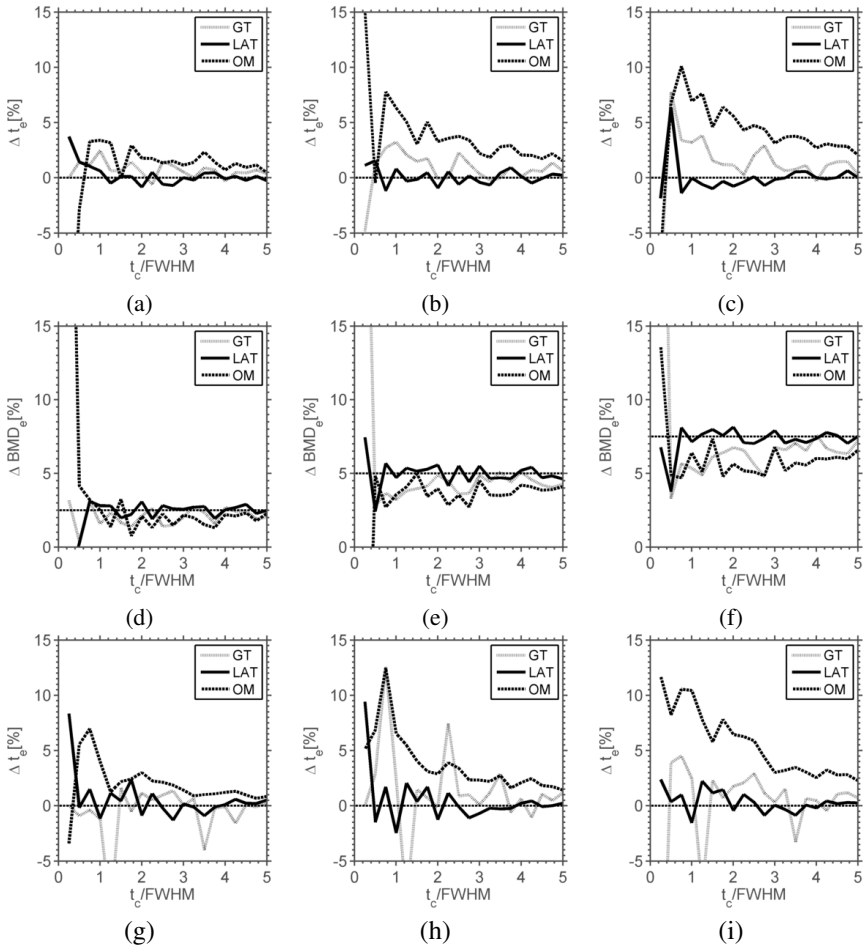
With LAT, the increase of cortical thickness is underestimated for thin cortices, but this method provides a good accuracy for $t_c > 2$ FWHM, whereas GT leads to an underestimation of Δt_c even for $t_c = 3$ FWHM. OM shows the best results in particular in the range FWHM <

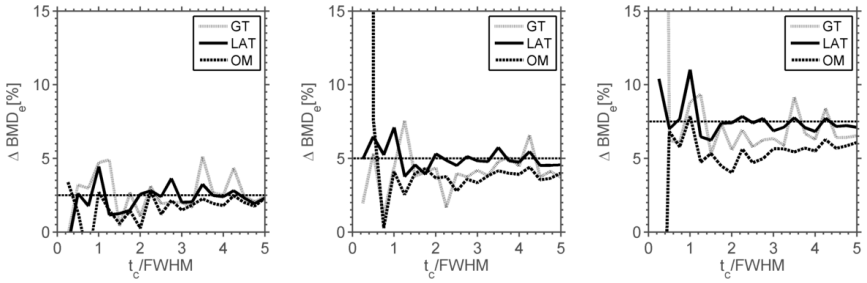
$t_c < 2$ FWHM. For smaller t_c, OM is severely impacted by large variances. For all three segmentation techniques, the assumed increase in cortical thickness results in an artificial increase of cortical BMD, which was larger for thinner cortices.

3.2 Variation of Cortical BMD

The effects of an assumed longitudinal 2.5 %, 5.0 % and 7.5 % cortical BMD increase on measured changes (Δt_e and ΔBMD_e) at two different noise levels are shown in Fig. 3 as a function of t_c/FWHM.

With OM, the simulated increase in BMD_c results in a falsely detected increase of Δt_e, which was larger for thinner cortices, and an underestimation of ΔBMD_c even for $t_c = 4$ FWHM. It must be remembered that BMD_c used in the fit is set to 1400 mg/cm^3 and is not adapted to the simulated BMD_c change. Furthermore, it can be questioned whether a 5 % change at the location of interest also occurs in the region where the true value is determined. The use of LAT shows small changes in Δt_e and a slight overestimation of ΔBMD_c for $t_c < 2$ FWHM.





(k) (l) (m)

Figure 3: **Increase of BMD_c.** Mean values of estimated changes of Dte ((a) to (c) and (g) to (i)) and ΔBMD_e ((d) to (f) and (k) to (m)) as a function of $t_c/FWHM$ for an assumed 2.5 % (first column 1), 5. % (second column) and 7.5 % (third column) increase of true cortical BMCc. (a) to (f): $\sigma_{noise} = 30\ mg/cm^3$, (g) to (m): $\sigma_{noise} = 37\ mg/cm^3$.

4 Discussion

Three different segmentation techniques were used to quantify longitudinal changes of cortical thickness and BMD. All simulations were performed for two different noise levels.

For the lower noise level each algorithm show good results for $t_c > 2$ FWHM. For thinner cortices, OM performs best in detecting changes of t_c. All three methods, however, overestimate ΔBMD_c for $t_c < 2$ FWHM. A true change in cortical BMD with constant cortical thickness can most accurately be measured with LAT and GT for $t_c >$ FWHM. OM underestimates ΔBMD_c, and measures a false increase in cortical thickness, which is not the case for LAT and GT. For $t_c <$ FWHM all three segmentation algorithms are strongly affected by increasing noise in particular with respect to ΔBMD_e. As a consequence, changes of cortical thickness and BMD are much harder to detect even in the range of $t_c < 3$ FWHM.

These results must still be verified in more advanced simulations, e. g. considering periosteal apposition.

Acknowledgements

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