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NOTE

An anatomically shaped lower body model for CT scanning of cadaver femurs

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Abstract

Bone specific, CT-based finite element (FE) analyses have great potential to accurately predict the fracture risk of deteriorated bones. However, it has been shown that differences exist between FE-models of femora scanned in a water basin or scanned *in situ* within the human body, as caused by differences in measured bone mineral densities (BMD). In this study we hypothesized that these differences can be reduced by re-creating the patient CT-conditions by using an anatomically shaped physical model of the lower body. BMD distributions were obtained from four different femora that were scanned under three conditions: (1) *in situ* within the cadaver body, (2) in a water basin and (3) in the body model. The BMD of the three scanning protocols were compared at two locations: proximally, in the trabecular bone of the femoral head, and in the cortical bone of the femoral shaft. Proximally, no significant differences in BMD were found between the *in situ* scans and the scans in the body model, whereas the densities from the water basin scans were on average 10.8% lower than *in situ*. In the femoral shaft the differences between the three scanning protocols were insignificant. In conclusion, the body model better approached the *in situ* situation than a water basin. Future studies can use this body model to mimic patient situations and to develop protocols to improve the performance of the FE-models in actual patients.

(Some figures in this article are in colour only in the electronic version)

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Introduction

Reliable prediction of fracture risk is extremely important to select the correct treatment protocol for patients suffering osteoporosis or metastasized cancer in the skeleton.

Clinically, bone mineral density (BMD) measurements are commonly used to estimate fracture risk in osteoporotic patients, but this method has poor sensitivity and specificity (Melton *et al* 1993, Rüegsegger 1991, Small 2005, Tanck *et al* 2009a).

For patients with femoral bone metastases it is very difficult to determine the fracture risk based on imaging data (x-ray or CT), even for experienced physicians. As a result, many patients are surgically over-treated whereas some patients, who are defined to be at low risk, fracture their bones (Van der Linden *et al* 2003, 2004). Hence, for both metastatic and osteoporotic patients, the fracture risk assessment methods that are clinically used are currently unsatisfying.

Bone specific, CT-based finite element (FE) analyses have great potential to improve the prediction of fracture risk. The FE-method with nonlinear mechanical properties has shown to be very promising in quantifying bone strengths under compression and high correlations, of up to $r^2 = 0.96$, were found between the failure force in laboratory experiments and FE predictions (Bessho *et al* 2007, Keyak *et al* 2005, Tanck *et al* 2009b).

The FE-method in actual patients has probably lower performance. From quantitative CT, the FE-models derive the bone geometry and the BMD distribution, which determines the mechanical behavior of the model (Keyak *et al* 2005, Taddei *et al* 2006). It should be noted that BMD as measured with the CT scanner is only an approximation of the apparent bone volume fraction; this method is too coarse to capture the tissue properties on a trabecular scale. The BMD is obtained from the CT values using a calibration phantom with known densities. However, the BMD is influenced by several factors, including the applied CT scanning energy, the scanner itself, the location of the calibration phantom and the volume of the scanned object (Cann 1988), but also by the fact whether the bones are individually scanned (bare femur in a water basin) or scanned *in situ* within the human body. Keyak and Falkinstein (2003) showed in a pilot study that the latter factor could lead to a difference of 13% in fracture strength prediction using the FE-method. This difference is related to volume effects, beam hardening and scatter effects, the correction of which by the scanner software is almost perfect in a water geometry, but causes artifacts when non-water tissues are present. Hence, in a water basin with bare femurs, these effects are minor compared to the *in situ* situation, in which the contra-lateral femur, the pelvic bone and soft tissue structures are present. Hence, when the FE-method is directly applied to patients, it is likely that errors will occur as the *in situ* environment of a patient's femur is more complex than the environment of a femur in *in vitro* laboratory experiments. Besides the application to fracture prediction based on the FE-method, this study applies to fracture prediction based on BMD as well.

In this study we hypothesized that the differences between *in situ* and *in vitro* can be reduced by re-creating the patient CT-conditions by using an anatomically shaped physical model of the lower body. It is hypothesized that the body model will produce similar BMD estimates as *in situ* and better approaches the *in situ* situation than a water basin.

Material and methods

Development of the anatomically shaped model of the lower body

The dimensions of the anatomically shaped model of a lower body were based on mean dimensions from Dutch males and females between 20 and 60 years (www.DINed.nl). From

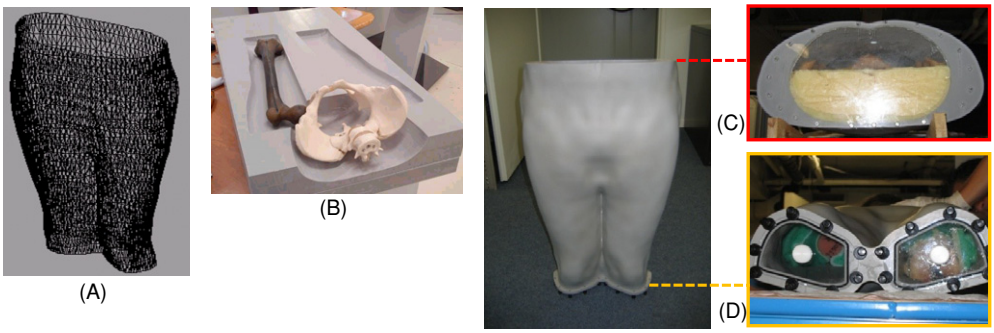

Figure 1. Development of the anatomically shaped model of the lower body. (A) The scaled contour of the body model; (B) within the model, a cavity was created for a pelvis and two femurs; (C) at the proximal side, the model was semi-permanently closed; (D) at the distal side, two removable Lucite plates containing air valves were incorporated.

an *in vivo* CT-scan of a patient the contours of the lower body were extracted, from belly-button to the knees, and scaled to the right Caucasian dimensions (figure 1(A)). Main dimensions were hip breadth 0.394 m, hip circumference 1.044 m and thigh thickness 0.149 (www.DINed.nl).

Based on the scaled contour, a body model with a polystyrene wall was produced (Hemabo, Enschede, The Netherlands). Polystyrene was chosen as it has an x-ray attenuation coefficient comparable to soft tissue material. Within the model, a cavity was created for a human pelvis and two femurs (figure 1(B)). The length of the inside femur cavity was 0.54 m, and the width at the femur condyles was 0.11 m.

A pelvis was harvested from a medium cadaver body and subsequently fixated in formalin. To secure the pelvic bone within the body model, melted synthetic wax was used, which renders a water equivalent CT-scan image. At the proximal side, the model was semi-permanently closed by a Lucite plate with screws (figure 1(C)). The positioning of the femurs and the filling of the remaining spaces with water (to simulate the remaining volume of soft tissue surrounding the bone) occurred at the distal end of the model. At this end the model was equipped with two removable Lucite plates containing air valves (figure 1(D)). The valves were added to ensure that no air would be entrapped as it is known that air pockets affect CT-densities (Cann 1988). With this set-up, femurs could be placed in the anatomical position.

Specimen

Four human cadavers were selected for this study, three males and one female aged between 69 and 87. To obtain a variety in body size and weight, the four cadavers were visually selected on body size, which was defined as being small, medium, large, and extra large.

The femur region of each cadaver body was CT-scanned (Philips, single-slice ACQsim) at the following settings: 120 kV; 220 mAs; 3 mm slices; scan field of view 48 cm; pitch 1.5; spiral; pixel matrix 512 × 512; pixel size 0.9375. Subsequently, from each cadaver, one stripped femur was obtained and scanned in a Lucite water basin (dimensions (w × h × l) 15 cm × 20 cm × 60 cm) with a 15 cm water level. The same femur was also scanned in the anatomically shaped physical model of the lower body (body model). Hence, each femur was scanned using three different scanning methods: *in situ*, in a water basin, and in the body model (figure 2). All scans were made on top of a solid calibration phantom (Image Analysis), which contained rods of calcium hydroxyapatite in tissue-equivalent material. The

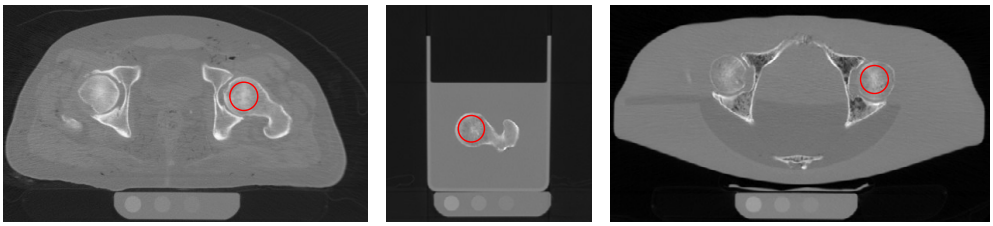

Figure 2. The three different scanning methods for each femur, from left to right: *in situ*, in a water basin and in the body model. The red circle is part of the proximal volume of interest in the trabecular bone of the femoral head.

known concentrations of calcium hydroxyapatite in the rods were 0, 50, 100, and 200 mg ml^{-1}. The BMD values of the voxels were calculated using the calibration values within the same CT-scan slice (Binkowski *et al* 2008). Hence, a slice-specific calibration method was applied.

Analyses

To enable comparison of the BMD values obtained in the three data sets the 3-dimensional (3D) images were registered relative to each other. For this purpose in-house developed registration software (Fusion 1.0.12, Department of Radiology) was used. The three sets of CT-scans were registered with a precision of 0.1° in rotation around three axes and 0.1 mm in translation along the three anatomical axes. After registration, a spherical volume of interest (VOI) was chosen proximally, in the trabecular bone of the femoral head (figure 2). A second VOI was constructed in the cortical bone of the femoral shaft, covering 20 consecutive slices approximately 15–20 cm below the femoral head.

Subsequently, the mean BMDs of the *in situ* scans, the water basin scans and the body model scans were statistically compared. For this, we built a linear mixed model in a statistical program, i.e. SPSS for windows (SPSS Inc, release 14.0), with the three scanning methods 'water basin, in situ scan and body model' as fixed factors and BMD as a variable. A two-sided significance level (alfa 0.05) was applied.

Results

For the VOI in the femoral head, no significant differences in BMD were found between the *in situ* scans and the scans in the body model, whereas the BMD from the water basin scans was on average 10.8% lower than the ones scanned *in situ* (range 6.4–15.3%) ($p < 0.001$) (figure 3). In three out of four measured bones the body model BMD reproduced the *in situ* BMD very well; in the smallest cadaver, the body model produced a lower BMD value similar to the water basin scan.

For the VOI in the femoral shaft, the relative differences between the three methods were smaller than those obtained for the VOI in the femoral head. The difference in BMD between the water basin scan and *in situ* situation was 2.5% (range 0.9–4.0%); between *in situ* and lower body model 3.3% (range 2.4–5.0%) and between lower body model and water basin 1.3% (range 0.4–2.7%) (figure 3). In this VOI, no significant differences were found between the three methods.

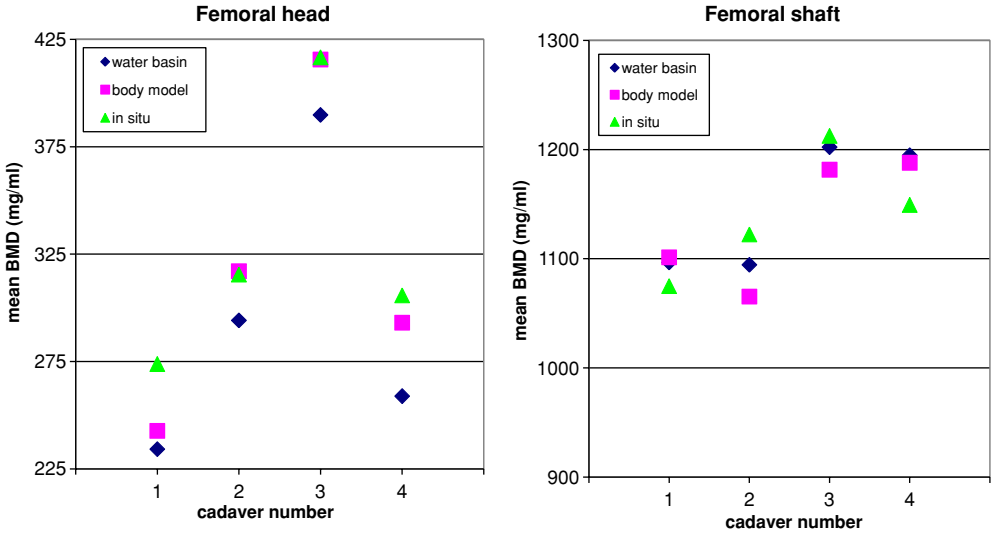

Figure 3. The mean BMD for the three scanning methods in the trabecular bone of the femoral head (left) and the cortical bone of the femoral shaft (right) for each of the four specimens.

Discussion

This study has shown that using CT-scan data obtained from a water basin or *in situ* situation can have a substantial effect on BMD, with differences up to 15% in the proximal femur, where most fractures occur. A small deviation in density may result in a much larger deviation in the related mechanical properties (Keyak *et al* 2005, Taddei *et al* 2006), thereby considerably affecting the FE fracture load prediction. The parameter settings of FE-simulations based on water basin CT-scanning protocols are, therefore, not necessarily valid for patient specific studies. In future plans the real effect on the FE fracture load will be analyzed by generating FE-models from data under the three scanning conditions. The FE-models will be loaded to failure and the failure loads will be compared to each other.

In the cortical bone of the femoral shaft, differences in density were relatively small between the scanning methods. Hence, the water basin can mimic the soft tissues surrounding an *in situ* femur shaft rather well; in both situations, the volume around the femur shaft and the tissues that cause beam hardening are rather similar, which resulted in similar attenuation of the x-ray, hence similar BMDs. Proximally, however, the volume effect and beam hardening effect are a source of error due to the presence of the pelvis, contralateral femur and soft tissue, which are not present in the water basin. In the case of the water bath (where no pelvic bone was apparent) the Hounsfield units of the calibration phantom were rather constant in each CT slice (data not shown). However, the Hounsfield units of the calibration phantom varied considerably in cases where the pelvic bone was present (the *in situ* and body phantom cases), indicating that the presence of the pelvic bone had a considerable effect on the CT attenuation in this region. This results in different x-ray attenuations, hence BMDs, between the *in situ* and water basin scans. This study showed that by using an anatomically shaped model of the lower body these differences can be reduced.

A limitation of the anatomically shaped model is its fixed dimensions. Therefore, the body model not always perfectly matches the *in situ* situation. Bone and soft tissue quality and dimensions affect the beam hardening effects. This was shown by the results of cadaver

1, the smallest sized specimen, of which the dimensions were most differently from the body model. When scanned in the body model, the femoral head density of specimen 1 was slightly closer to the *in situ* situation than the water basin was, but the results were not as good as the three cadavers sized from medium to extra large (figure 3). Further analyses are required to confirm that body size was indeed the crucial factor or if other unknown factors were involved.

Our ultimate goal is to be able to measure the structural geometry of a patient using CT then map the CT data to a finite element (FE) model to calculate bone strength. A common method for validating these models is to scan a cadaver femur ex vivo then compare the FE strength prediction to that measured experimentally. Our physical model of the body provides more realistic scanning conditions for cadaver femurs to meet our ultimate goal. In addition, the use of the body phantom may enable us to improve and choose the best calibration procedure in patients.

In conclusion, the body model better approached the *in situ* situation than a water basin, particularly in the proximal area where femurs usually fail. Future studies can use this body model to mimic patient situations and to develop protocols to improve the performance of the FE-models in actual patients.

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