Computational analysis of the effect of physical activity on
the changes in femoral bone density in prepubertal
children

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Abstract

Human bone health, and osteoporosis in particular, are a highly topical research. A
causal and positive effect of physical activity on bone mineral density is supported by
a bunch of clinical-based research studies. Moreover, actual research suggests long-term
benefits of childhood physical activity to the prevention of osteoporosis in adulthood. Com-
plementary to the clinical-based study is the computational analysis, based on a bone re-
modeling model. Although many different remodeling models are developed by several
research groups, very little research is done in the application of computational remodeling
in children. The current study is performed in this gap and consists of two major parts.
First, a model is developed to represent the femoral bone of a prepubertal child. A finite
element with assigned material properties based on Computed Tomography scans is created
and verified. The musculoskeletal loads are obtained from experimental kinematic data by
the faculty of movement. Gait, stance, run and countermovement jump as performed by
prepubertal children are represented by superposition of crucial timeframes in the activity
cycle. The effect of physical activity on bone mineral density in prepubertal children is
investigated with the Lisbon bone remodeling model. An optimal value of model parame-
ters $k$ and $m$ is investigated, which take into account the inter-subject variability. A value
of $k = 0.0025$ and $m = 4$ are considered as best to represent the prepubertal population.

In the second part of the research, the effect of additional physical activity on bone min-
eral density is investigated. It is concluded that results of the computational model are in
agreement with clinical results. Spending more time in physical activity, high-intensive ac-
tivity in particular, the bone mineral density increases. It is stated that an initial increase
in activity time leads to a greater increase in bone mass compared to a further increase
in activity time. Further, the results suggest that physical activity decreases the risk for
typical (osteoporotic) femoral fractures, as suggested by former research. In the region of
the femoral neck, this happens by the periosteal apposition of bone: a relative increase in
density of the outer shell. In the trochanteric region, the fracture risk is decreased by an
increased bone mineral density.

1 Introduction

Osteoporosis is a musculoskeletal disease characterized by reduced bone mineral density (BMD)
and increased risk of fragility fractures. A vulnerable region for decreased density is the proximal
femur. Osteoporotic fractures result in significant mortality and morbidity and lead to consid-
erable societal costs, including direct medical costs and indirect costs resulting from reduced
quality of life, disability, and death. The International Osteoporosis Foundation proposed that
prevention is the best method to fight against this disease. An important marker of skeletal

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health is peak bone mass. This is considered to be a strong predictor of osteoporosis risk in later life.

The bone mass of the human skeleton changes through the concept of bone remodeling. This is a continuous process of replacing bone for maintaining normal calcium levels in the body and to maintain bone strength. The outcome of the process is the replacement of a small amount of the pre-existing bone by a comparable, a larger or a smaller piece of new bone. Mechanical loading of the bone affect the bone remodeling process and promotes increased bone mass. This mechanical loading happens during weight-bearing physical activity.

The growing skeleton has a greater capacity to adapt to loads associated with weight-bearing activity than the mature skeleton [1]. These findings suggest that the years of childhood and adolescence represent an opportune period during which bone adapts particularly efficiently to such loading: a ‘window of opportunity’. The effect of physical activity on bone health is extensively investigated based on clinical results. Despite the great benefits of these studies, they are limited to a time scale (often several month between two measurements) and to existing medical equipment to measure density (rarely volumetric bone density). A computational analysis gives a complementary view to obtain more complete data.

Together, these findings strengthen the notion that maximizing bone health in the proximal femur during childhood growth, especially during prepubertal years, may represent an important strategy in the prevention of osteoporosis and fractures of the femur during ageing.

2 Methods

2.1 Lisbon Model

The bone remodeling process is described in several computational models. This study makes use of the Lisbon model, developed by Fernades et al [2]. It is based on Wolff’s law and assumes that bone is able to adapt itself in order to attain the stiffest structure for a given set of loads, while taking into account biological cost parameters that controls total bone mass.

On a microscale, the model describes the unit cel as a cubic with a prismatic hole. The dimensions of the hole are given by the parameters \(a_1\), \(a_2\) and \(a_3\) and the orientation of the unit cell is given by the Euler angles \(\theta = [\theta_1 \theta_2 \theta_3]^{-1}\). The relative density of the unit cell is stated as \(\mu = 1 - a_1 a_2 a_3\). A periodic repetition of the unit cell leads to a model description on macroscale, where the model considers bone tissue as a linearly elastic orthotropic material. The remodeling law is stated as:

\[
\sum_{p=1}^{NC} \left( \alpha_p \frac{\partial E^H}{\partial a} e_{ijkl} (u^p) e_{ij} (u^p) \right) - k \frac{\partial \mu^m}{\partial a} = 0 \tag{1}
\]

The first term is the mechanical advantage, constricted to the metabolic cost represented by the second term. \(E\) are the homogenized elastic properties. Components of the strain fields \(e_{ij}\) en \(e_{ijkl}\) are dependent of the set of displacement fields up. The mechanical advantage (first term) is the sum over the \(p\) loading conditions. Each loading condition has a weigh \(\alpha_p\). The metabolic cost is dependent of the change of relative density \(\mu\). \(k\) and \(m\) are parameters dependent on the individual. Model parameter \(a\) is considered constant within each finite element.

2.2 Creation of a prepubertal model of the femur

Prediction of the bone remodeling process of prepubertal children, requires a subject specific model and boundary conditions.

2.2.1 Finite element model

The geometry of the proximal femur is created based on medical CT-scans with Mimics software (Materialise NV, Leuven) for segmentation and pre-processing. A 3D model is created based on the segmented images. Next, a surface mesh of triangular elements and nodes of the 3D
model is generated with 3-Matic software (Materialise NV, Leuven). Further pre-processing is done including smoothing, reducing the number of elements and optimizing the quality of the triangles. The result is imported into SolidWorks (Dassault Systèmes, Velizy-Villacoublay) as a pointcloud file making use of the Scanto3D Module. A slice of 1 mm of the distal femur is cut off to obtain a flat surface and a solid model is created. Finally, the model is imported into Abaqus (Dassault Systèmes, Velizy-Villacoublay) as a Parasolid file, where a volume mesh is obtained. The mesh consists of C3D4 elements, which has the advantage of an automated meshing approach. Moreover, these elements allow for results more closely to theoretical ones for the simplified proximal femur compared to other element types according to Ramos et al [3]. Small geometric features are assigned a finer mesh. To ensure accurate numeric results, mesh convergence tests are performed. Convergence is checked using three different solution variables. Maximum displacement, maximum Von Mises stress and total strain energy. The optimal volume mesh consists of 28732 nodes and 145051 elements.

Material properties are assigned to the model by importing the finite element meshes into Mimics again. Densities are assigned to elements by relating the grayvalues from the CT-image to apparent bone mineral density. A linear relationship is assumed between high porotic trabecular bone (0.1 g/cm$^3$) and cortical bone in the femoral shaft of a child (1.3 g/cm$^3$). Calibrating these points with the respective grayvalues of 1133 and 2813 for the two types of tissue, leads to the following relationship:

$$\mu = -0.5137 + 0.0005467 GV \quad (2)$$

Blurring of borders between cortical and trabecular bone occurs, as one element of the model represents several pixels of the CT-scan. This small error could be avoided with a finer mesh and an average edge length similar to the CT in-plane resolution. However, this is computational not advantageous. The limits of the density range are set to 0.056 g/cm$^3$ and 1.1 g/cm$^3$. 2.5% of the elements falling outside this range are corrected, in order to avoid biologically unrealistic densities. The bone material was considered to be linear elastic with a Young’s modulus of 20 GPa and a Poisson’s ratio of 0.3 for the base material.

To avoid a checkerboard density patter, an optimization for elements is preferred. The assigned material properties are of the elements interpolated to the nodes. The density of each node is defined as a weighted average of the surrounding elements with the volume of the elements as weights.

The proximal femur is constrained at the bottom surface of the cutted shaft. Some experiments pointed out that the condition of all nodes of the ground surface fully constrained in three directions avoids best physiologically unrealistic stress concentrations.

### 2.2.2 Musculoskeletal loading

The loading configuration of the FE model does play an important role in the outcome of the remodeling simulation. A description of the physiological muscle loading that is as complete as possible is included, based on experimental values calculated by Professor Filipa João of the Faculty of Human Kinetics, University of Lisbon. In the experiments participates a typically developing male subject of the aimed population.

3D gait analysis, recording kinematics and ground reaction forces are performed to provide the input experimental data for the musculoskeletal model. The musculoskeletal model OpenSim v3.2 presented by Delp et al [4] and a scaled version of the gait 2392 computer model are used in the calculation of kinematics and inverse dynamics. Joint moments are calculated with the inverse dynamics tool in OpenSim, based on the respective ground reaction forces and gait kinematics. A static optimization algorithm in OpenSim is used to estimate the underlying muscle forces required to balance the joint moments, while minimizing the global amount of muscle activation. To calculate the resultant joint loads, the JointReaction algorithm in OpenSim is used.

Only relevant muscles for bone remodeling of the proximal femur are taken into account into the calculations. Those are: Gluteus Minimus 1, Gluteus Minimus 2, Gluteus Minimus 3, Gluteus Medius 1, Gluteus Medius 2, Gluteus Medius 3, Gluteus Maximus 1, Gluteus Maximus 2,
Iliacus, Psoas, Quadratus Femoris, Gemellus, Piriformis, Pectineus and Adductor Brevis. Four types of physical activities are performed in the experiments: gait, stance, countermovement jump and run. To obtain a representation of the complete activity cycle, physiological loading is approximated with a superposition of different loads of different timeframes of the activity cycle. For one activity cycle, timeframes are selected that represent a muscle and hip contact loading with a (local) maximum in ground reaction force or a difference in direction of the hip contact force. The resulting timeframes of the cycles are are listed in table 1.

<table>
<thead>
<tr>
<th>Activity</th>
<th>Label</th>
<th>Timeframe</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gait</td>
<td>gait1</td>
<td>Maximal Weight acceptance</td>
</tr>
<tr>
<td></td>
<td>gait2</td>
<td>Midstance</td>
</tr>
<tr>
<td></td>
<td>gait3</td>
<td>Push-off</td>
</tr>
<tr>
<td>Stance</td>
<td>stance</td>
<td>Balanced standing</td>
</tr>
<tr>
<td>Jump</td>
<td>jump1</td>
<td>Take off</td>
</tr>
<tr>
<td></td>
<td>jump2</td>
<td>Landing</td>
</tr>
<tr>
<td>Run</td>
<td>run1</td>
<td>Impact peak</td>
</tr>
<tr>
<td></td>
<td>run2</td>
<td>Active peak</td>
</tr>
<tr>
<td></td>
<td>run3</td>
<td>Push-off</td>
</tr>
</tbody>
</table>

Table 1: Types of physical activity and the selected timeframes.

The direction of the muscle force or 'line of action', is calculated as the connection of two points. The start and end point are extracted of OpenSim. Muscle attachments are extracted from OpenSim, which is defined as the location of the mechanical centroid of muscle lines of action. The shape of the femur in OpenSim and the finite element model are different. To match the muscle attachments, a transformation matrix is built which compensates for the difference in shape. This is done by selecting six equal landmarks on both the femur of OpenSim and the finite element model. These landmarks are matched with an Iterative Closest Point algorithm. This results in the following matrix:

\[
\begin{bmatrix}
0.997 & 0.0378 & 0.0624 & -18.52 \\
0.0358 & 0.0999 & 0.0328 & 6.88 \\
0.0636 & 0.0305 & 0.998 & -2.94 \\
0 & 0 & 0 & 1
\end{bmatrix}
\]

A muscle attachment point on the finite element model is obtained by expressing the coordinates in the coordinate system attached to the femur, multiplying it with the matrix above to compensate for the shape and finally projecting it on the surface. To mimic physiological loading conditions, a muscle attachment area instead of a point is defined. The area of muscle attachment is defined with a manual procedure based on previously measured physiological muscle attachment areas. The loads of the experimental data are applied as a concentrated load at the calculated point of attachment. To distribute the load over the muscle area, the nodes of the selected surface are coupled in Abaqus. A surface-based coupling, couples the motion of a collection of nodes on a surface to the motion of a reference point. The distributing coupling is used, so that the motion of the surface nodes is constrained to the translation and rotation of the reference point. The constraint distributes loads such that the resultants of the forces (and moments) at the coupling nodes are equivalent to the forces and moments at the reference node.

The hip and the femur make contact through a cartilage layer in the hip joint. The anatomical contact area is defined by thresholding the maximal distance between femoral head and hip and equals $1575mm^2$. As the contact area changes during movement, a portion of the the anatomical maximum contact area is calculated, according to Yoshida [5]. For walking and running, 79% or $1244mm^2$ is estimated. For jump, 52% or $819mm^2$ is estimated. The hip contact load is applied to the finite element model on a similar way as the muscle load. First, the point of peak load is calculated. The defined surface is coupled to the point of peak load with a distributing coupling. However, different to the muscle loading, the hip contact load is not uniform. It decreases from the pole...
(point of peak load) to the outer area. Tests are performed with different weighting functions. The result of these tests is compared qualitatively with literature about the pressure distribution. Of the linear, the quadratic and a cubic weighting scheme, the cubic is considered best, which is shown in figure 1.

Figure 1: Stress distributions at the femoral head due to hip contact force. Rainbowspectrum red (highest) to blue (lowest) stress. (a) Result as calculated by Yoshida [5]. (b) Calculated result with cubic distributed coupled surface.

Several parameters characteristic for the Lisbon remodeling model need to be adjusted to represent the prepubertal population. The model is calibrated to this inter individual biological variability by means of the parameters $k$ and $m$, which represent the cost of maintaining bone. This is done by simulating the activity of a non-active subject whose activity consist of walking only. A uniform initial density equal to the median of the CT densities is assumed. This impartial value is selected to allow each element to change to lower or higher densities with minimal cost. The algorithm is run for different parameters $k$ and $m$. The resulting density pattern of each combination of $k$ and $m$ is compared to the real densities of the medical scan. The best match between the densities indicates an optimal combination of parameters $k$ and $m$ for the subject. The comparison between the real density values (scan) and the calculated values (model) is based on a statistical analysis which takes into account the mean, the standard deviation, a correlation coefficient, the absolute error, the relative error and an average error which is a weighted average of the absolute and relative error.

Further, parameter $\alpha^p$ is adjusted. It is the load weight factor and indicates the relative frequency of each type of physical activity. The sum of all $\alpha^p$ is defined as $\alpha$, which has a maximal value of $\alpha = 1$. This parameter is coupled to time spent in physical activity. The total time spent in physical activity by prepubertal children is estimated as approximately 400 minutes per day which corresponds to $\alpha = 1$. A value of $\alpha^p$ per minute can be defined as $\alpha^p = 1/400 = 0.0025 \text{min}^{-1}$. The total time $\alpha$, could be divided over several types of physical activity. For each type of activity, it is assumed that each timeframe has an equal contribution. If the subject performs less activity than 400 min, this is considered as the activity rest.

2.3 Change in physical activity

To determine a change in density due to physical activity, a reference group is created. The reference group performs only the baseline level of physical activity. It is the activity of an average non-active child whose physical activity consists of walking, standing and rest. This child is the reference group. It is assumed that as much time walking is spend as standing. This baseline level of activity is defined as the activity which establishes a femoral neck density equal to a value found in literature of $0.393 \text{g/cm}^3$ [6]. Two reference groups are defined which spend additional time in physical activity above the baseline amount. The additional physical activity consisted of increments of 5 minutes jumping and 5 minutes run. This combination of run and jump represents children’s activity, which includes many different movements running, hopping, climbing, jumping etc. The second reference group spends the same additional activity time, but spends it walking.

Several regions of interest (ROI) are selected to investigate the change in bone density. Four standard ROIs and three additional ROIs (total of seven ROIs) are selected as proposed by Prevrhal et al [7]. The additional defined ROI have the advantage to be able to differentiate changes in trabecular and cortical bone density at the proximal femur. The ROI are shown in figure 2.
3 Results and discussion

The thesis consisted of two main goals: the creation of a femoral model of a prepubertal child and investigating the changes in density with respect to physical activity.

3.1 Model of the prepubertal femur

In children’s bone three processes are active: growth, modeling and remodeling. A phenomenological model as the Lisbon model is able to deal with all three processes at once, since there is no relationship between biology and modelship. Further, the Lisbon model has the variable cost parameters \( k \) and \( m \). These parameters define the cost of bone maintenance and therefore, control the total amount of bone mass. They depend on several biological factors such as gender, age, hormonal status and disease. In this way, the model captures the possible age or growth related differences.

From major importance is the subject specificity of the model. The geometry is created based on the CT scans with standard segmentation protocols as described by Materialise in the Mimics reference guide. A qualitative evaluation of the geometry and the material assignment is done by comparing a virtual X-ray of the model with a DEXA scan of clinical practice (figure 3). The cortical shell on the femoral shaft is clearly visible on both the images. Moreover, a dark area of low density is noticed in the trochanteric region. A possible critique is the apparent circulars line around the femoral head, which is possible an inaccuracy of the segmentation. Because no region of interest is focusing on the femoral head, this is only a minor concern.

A quantitative analysis could be done based on the characteristics of the selected ROIs. Table 2 lists the densities at the ROI measured from the model. These values only report a starting pattern and are not yet the densities of the reference group, since no baseline activity is applied yet. Ward triangle is per definition an area of diminished density. The value is indeed lower than the other ROI except for the CIRCROI and the TROCH. As the CIRCROI and CENTRAL are focusing on the trabecular rich regions, the low values are explained. Further,
the trochanteric region is also lower than the average due to the amount of porotic trabecular bone. CORTROI on the other hand, focuses on the weight bearing cortical bone of the femoral shaft, which justifies the high density. TOT FEM is a good reference for the average density of the proximal femur. It can be concluded that the relative densities of the different ROI are correct representation. Speaking in absolute terms, the densities of trabecular and cortical bone are calibrated with values listed in peer-reviewed research. Moreover, the amount of baseline activity of the reference group is defined as the activity which establishes a femoral neck density equal to the value found in literature.

<table>
<thead>
<tr>
<th>ROI</th>
<th>Density [g/cm$^3$]</th>
</tr>
</thead>
<tbody>
<tr>
<td>CENTRAL</td>
<td>0.283</td>
</tr>
<tr>
<td>CIRCROI</td>
<td>0.258</td>
</tr>
<tr>
<td>CORTROI</td>
<td>0.526</td>
</tr>
<tr>
<td>NECK</td>
<td>0.304</td>
</tr>
<tr>
<td>TOT FEM</td>
<td>0.324</td>
</tr>
<tr>
<td>TROCH</td>
<td>0.231</td>
</tr>
<tr>
<td>WARD</td>
<td>0.260</td>
</tr>
</tbody>
</table>

Table 2: Starting densities in the different regions of interest as measured on the created model.

As parameters $k$ and $m$ determine the intersubject variability, an accurate selection of these parameters is important. In particular, because this is how the Lisbon model is set to remodel bone of children. The correlation coefficient and the average error show opposing tendencies. A trade-off between all tendencies should be made. Parameters $k = 0.0025$ and $m = 4$ are considered best, especially because it has only a very small difference with the mean ($\Delta\text{mean} = 0.034$) and standard deviation ($\Delta SD = 0.0231$). For the chosen value of parameters $k$ and $m$, a correlation coefficient of 0.666 is obtained. The average error, which is a relative value, is 0.274. All the tested statistical parameters give in general better results compared to Sharma [9] et al or Quental [10], who did similar analyses for the scapula. This could be explained by the simpler density distribution of the femur compared to the scapula.

Despite the try to underpin each assumption, there are some limitations or potential errors in the model. First of all, geometric adaptations, which may occur more readily in children, are not taken into account. On the other hand, the geometric adaptation is especially important when one would investigate the gain in strength due to physical activity. Secondly, the selected timeframes are only a representation of the activity cycle. Further, the musculoskeletal data and the CT images are from different subjects. Although they have very similar characteristics (sex, age, height and weight), there will be some slight intersubject variability. However, in this study there is no access to both kinematic data of movement analysis and CT scans from the same subject. More general, it is a limitation that only one subject is investigated to represent the average prepubertal child. As no standardized model of the prepubertal femur exist similar to the one created by Viceconti [11], this was the best solution. This is mainly a disadvantage when the results of this study would be compared with a similar study, but a different finite element model. Finally, factors that are not directly related to the mechanically induced bone remodeling, such as genetic predetermination, hormonal and central control of bone remodeling could skew the bone density predictions.

### 3.2 Change in physical activity

The results of the analysis are plotted in figure 4. The simulated data for intervention group 1 (increased high-intensive activity), intervention group 2 (increased time spent walking) are marked with a cross. Each data point represents the density of the ROI with respect to the additional amount of time per day spent on physical activity. The data are fitted with a smooth curve, to accentuate the tendency. Further, also the density of the reference group (only performing the baseline amount of activity) is plotted. Since the reference group does not perform any additional activity above the baseline activity, the curve is flat. Only the subregion femoral
The results for the other regions of interest have a very similar tendency and differ only in absolute terms.

Figure 4: Simulated data of density with respect to additional time spent in physical activity per day. Reference group 1 performs additional high-intensive activity. Intervention group 2 performs additional walking. The reference group does not perform additional activity.

These results are in accordance with clinical studies [12] examining physical activity levels and bone outcomes. This study convincingly shows that children who participate in higher levels of physical activity have greater bone mass accrual compared to less active children. This accounts for all investigated regions of interest. The effect of high-intensive activity is noticeable higher than the effect of walking. Sardinha et al [13] suggest that 25 minutes of activity are needed to improve remarkable benefits. The computational results show an effect on bone mass for each additional minute spent in physical activity. It is not known if the initial density accrual are not remarkable in clinical practice or that the initial density accrual is a bias of the remodeling model. Further investigation needs to point this out.

From the results, it is investigated how the density changes spending additional time in physical activity. In other words, the impact of an additional time increment on the density is investigated. From figure 4 it is seen that the fitted function is an increasing function. But, the increase becomes smaller when more time is spent in physical activity. This fact is verified by checking the derivative. The plot of figure 5 shows he derivative of the fitted function for intervention group 1 and 2 of the total proximal femur. The function indicates the immediate change in bone density with respect to changing time. All ROI show the same tendency, but for convenience only the total proximal femur is plot.

Figure 5: Change in bone density for additional time spent in physical activity.

The change in density is a decreasing function. This implicates that more bone mass is gained in with an initial increase in physical activity time. In other words, the bone has some resistance for adding more bone mineral content with increasing activity time. The derivative function seems to be convergent, so after a large amount of additional activity time bone changes linearly with time. This could be explained intuitively based on biological knowledge. As the tissue tends to be less porotic, or more mineral content is added, the additional mineral content comes with an increased metabolic cost. Looking at figure 5 at the derivative for intervention group
a rather flat curve is noticed. Indeed, the fitted function in figure 4 looks linear. Spending additional daily time walking results in a proportional increase of bone and activity time. As a conclusion, it is stated that an initial activity time increment leads to a greater increase in bone mass compared to another time increment. This only accounts for high-intensive activity and not for the gait cycle.

Some regions of interest are especially important since they are more vulnerable to fractures than others. Two types of fractures are frequently noticed: a fracture of the femoral neck and a fracture in the trochanteric region. It is noticed that physical activity decreases the risk for these typical (osteoporotic) fractures. In the trochanteric region, the fracture risk is decreased by an increased bone mineral density. In the region of the femoral neck, this happens also by the principle of periosteal apposition: a relative increase in density of the outer shell. This is inspected by investigating the ratio of the central region of interest to the femoral neck region. As all defined regions of interest have an increasing density with additional physical activity, the ratio is increasing as seen in figure 6. The lower subplot shows the derivative of the fitted function. As it is a decreasing function, this implicates that the central region grows less fast than the neck region, or more bone material is added to the outer shell than to the inner region.

![Figure 6: Upper: ratio of central to neck ROI. Lower: change in ratio with respect to increased activity. This indicates periosteal apposition. Plot for reference group 1.](image)

It is concluded that physical activity decreases the risk for typical (osteoporotic) femoral fractures. In the region of the femoral neck, this happens by the periosteal apposition: a relative increase in density of the outer shell. In the trochanteric region, the fracture risk is decreased by an increased bone mineral density.

### 4 Conclusion

As a first conclusion, it is said that the created model and loading is able to represent the prepubertal population. A finite element model is created and verified with a geometry and material properties of the prepubertal population. The musculoskeletal loading originates from experimental data. By including all relevant muscles and including several timeframes within one activity cycle, the physiological loading is more closely approximated.

The computational results agree with clinical-based research. Spending more time in physical activity per day, the bone mineral density increases. This accounts for all investigated regions of interest. The effect of high-intensive activity is remarkable higher than the effect of walking. Compared to the clinical-based research, the computational results are complementary and reveal more accurate information about the way how it changes and are able to differ easily between several regions of interest.

Further, it is stated that an initial activity time increment leads to a greater increase in bone mass compared to another time increment. This only accounts for high-intensive activity and
not for the gait cycle. Results suggest that low porotic bone has a higher resistance for added mineral content compared to high porotic bone.

As a last conclusion, it is suggested that physical activity decreases the risk for typical (osteoporotic) femoral fractures. In the region of the femoral neck, this happens by the periosteal apposition: a relative increase in density of the outer shell. In the trochanteric region, the fracture risk is decreased by an increased bone mineral density.

References


