Computational analysis of fractures in vertebral bodies

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Abstract

One of the diseases that affect many older people is osteoporosis and one of the major consequences of this disease is the vertebral body fracture, because osteoporosis is a disease where bone density decreases and the trabeculae became more thin and long and some of them disappear.

With these problems becomes essential to develop methods to better understand this disease and to prevent bone fractures, that is, the osteoporotic patient, with high probability of suffering a fracture, will be treated before the fracture occurs. The FE models arise as an opportunity that can significantly contribute to the development of methods that can help in medicine more precisely in this field.

A FE model based on CT scan images of a healthy human vertebrae was developed with the vertebral material properties defined using a bone remodelling model. The validation of these results was done through qualitative and quantitative comparison of FE models taken from literature.

Then, the simulation of different osteoporotic cases was performed and the comparison between them was made using a failure criterion for vertebral fractures based on the minimum principal strain. It was possible to verify that the most critical areas are the ones where the movements have higher probability to contribute to the vertebral body fractures and that abnormal movements influence the bone fractures.

Keywords:

Osteoporosis; Vertebral body fractures; FE Models; Bone remodelling model; Vertebroplasty and Kyphoplasty

1. Introduction

The motivation of this work comes from the necessity to answer to problems that result of the vertebral compression fracture. It is the most common clinical manifestation of osteoporosis which can cause a significant amount of pain. Bone mineral density (BMD) is what is mostly used in clinical environment to diagnosis osteoporosis, that is an indicator of vertebral fractures. But this method is not very efficient, so it became necessary find a more accurate way to predict osteoporosis and, more important, predict vertebral fractures.

Prediction is a very important tool for clinical application because if we can predict when and where the disease will appear it will revolutionise medicine. Because it will rise the probability of success in the intervention that the physician decides to apply to the patient to prevent the fracture and such intervention has an easier recover than a recover from a fracture. In the majority of the cases the fractures are not detected so the patient has a bigger risk to suffer from another one. This scenery has consequences more devastating for the patient and in some cases the pain is chronical.

An answer to these problems is use the better imagological tools that we have nowadays in assembly with finite element models. With this approach we will intervene earlier and decrease the probability of the fracture risk assessment for
the spine. With this solution we rise the life quality of the patient but also we can save money. Because the intervention is easier and the recovery to, so the patient goes home earlier, what is a positive point to the hospitals, and patient can return to his normal life and work.

So, this innovative method can be a revolution in clinical practice, change the patient life and can be a money saver for the hospitals and for the government. Nowadays with the boom of the technological innovation we have to explorer new approaches and try to ally medicine and technology. With this alliance the life quality of the patient can be improved significantly and the health gains are high.

2. Literature Review
Osteoporosis is a bone disease that affect the bone mass and structure, the bone mass decrease and the structure decays. Osteoporosis affect around 1 in 3 menopausal women's and 1 in 5 men, in total this disease affects 200 million people in the world.

Annually osteoporosis causes more than 8.9 million fractures worldwide and 50% of that fractures take place in Europe and in the USA. The fractures occur in different body parts, the majority of them in spine, hip and wrist. In Europe are estimated that in 2025 around 20 to 50% of the population with more than 65 years old will suffer a fracture due to the decrease of bone density.[1]

Osteoporotic fractures are deeply related with morbidity and mortality of the elderly population, as such this is an area of great concern, socially and economically.

About 25% of post-menopausal women suffer a vertebral fracture and these fractures could cause a chronic pain that is a disabling condition and the daily life of the person can be affected.[2]

In the European Union was estimated that in 2010 around 22 million women and 5.5 million men would have osteoporosis and in that year approximately 520 thousand vertebral fractures were registered. The economic impact of this pathology was estimated in a value around €37 billion whereby the incident fractures represent 66% of the costs.[3]

In 1994 the World Health Organization (WHO) published a report about the evaluation of the fractures risk and the application of these evaluation to diagnose osteoporosis, the criteria used was the measuring of the Bone Mineral Density (BMD). [4]

BMD is described in the majority of the cases as a T-score where the units are express in terms of standard deviation, in other words, T-score is the difference between the actual bone mass of an individual and the bone mass of the young healthy adults of the population, the bone mass peak for the healthy individuals.

For the evaluation of the BMD is necessary make a bone densitometry of the femur neck, then the value is evaluated and the WHO has defined that values lower than -2.5 SD means that the individual is osteoporotic. [3]

As we said earlier the major problem of osteoporosis is the bone fractures, in this work vertebral fractures are the issue we address, this type of fracture is treated with two kind of non-invasive methods: vertebroplasty and kyphoplasty.

But in most of the cases the physicians decided not to intervening internally in the patient and just give him drugs and the patient has to rest for a long period of time until the bone calcification. With this approach both the pain and the life quality doesn’t improve in the first month so this method isn’t the best approach.[5]

Vertebroplasty and kyphoplasty helps in the control of the pain but these two procedures have their risks and the application of those two procedures depending on the severity of the fracture, if the fractures is to serious is necessary invasive procedures.[6][7]

So ideally the best way is to prevent the fractures and the finite element models assembly with CT scan images can help in this field.

2.1 Bone Remodelling Model
The bone is a dynamic and active tissue, small changes in the bone architecture take place continuously. In the human skeleton occurs bone formation and bone reabsorption this process re known as bone remodelling.

In this study we used a bone remodelling bone developed by Fernandes et al. [8] on the basis of Wolff's Law.

The model in question combines thickness criteria with metabolic factors controlling bone mass. The bone is set as orthotropic material having a porous microstructure which allows different elastic properties in different directions.
The model microstructure is set as an open cell (unit), allows to do a simulation similar to the real bone, the cell is cubic and as dimensions, \( a_1, a_2, a_3 \). The relative density, \( \mu \), define the dimensions of the cell hole and is calculated by the next expression:

\[
\mu = 1 - a_1 a_2 - a_2 a_3 - a_1 a_3 + 2 a_1 a_2 a_3, \tag{1}
\]

para \( a_i \in [0,1] \)

Taking the above expression into account we have the bellow relations:

- For a null value of relative density, \( \mu = 0 \), that corresponds to the absence of bone in anatomic terms;
- For a unitary value of relative density, \( \mu = 1 \) (the maximum relative density), that corresponds to cortical bone;
- For a value between \( 0 < \mu < 1 \) of relative density, corresponds to trabecular bone.

The density distribution is obtained through an optimization problem defined by the maximization of structure stiffness. Since bone adapts in response to applied loads, the volume and consequently the bone mass will vary. This mass variation is taken in account introducing in the model an additional term in the objective function that depend on the total bone volume. This term can be seen as a metabolic cost to the body to keep the bone tissue where the metabolic factor is the parameter \( k \). If \( k \) increases, the metabolic cost is higher, and then the total bone mass decreases.

Using a multiple-load optimization criterion, the problem can be described by:

\[
\min \sum_{p=1}^{N_C} a^p \left( \int_{\Omega} f^p u^p_d dT + k \int_{\Omega} \mu(a)^m d\Omega \right) \tag{2}
\]

subjected to

\[
0 \leq a_i \leq 1, \quad i = 1, 2, 3 \tag{3}
\]

Where \( N_C \) is the number of possible cases for the applied load and \( a^p \) is the weight of each load which satisfy the following equation: \( \sum_{p=1}^{N_C} a^p = 1 \).

To solve the optimization problem given above a Lagrangian method is used, the optimum condition for the problem are given as:

\[
\sum_{p=1}^{N_C} a^p \left( \frac{\partial E_{ijkl}^H e_{ij}(u^p)}{\partial a} e_{ij}(u^p) \right) - k \frac{\partial \mu^m}{\partial a} = 0 \tag{4}
\]

which corresponds to the bone remodelling law. In equation (4) \( E_{ijkl}^H \) defines the homogenised material properties tensor, \( e_{ij} \) the strain field and \( u^p \) is the displacement field.

To solve the equations a numerical procedure is adopted as it possible see in the Figure 1.

![Figure 1 - Bone remodelling computational model. Adapted from [14].](image)

### 2.2 Failure Criterion

One of the main goals of this work was to try to see how it was possible to predict the failure of a vertebra, for such and after a thorough search in literature we realized that most of the studies treated bone as a fragile material (Chen et al. and Bessho et al.). This approach is made because bone have a lower resistance to tensile strength than compressive strength, therefore the bone is then treated as a brittle material.[9][10]

One of the criteria that could be used to predict vertebral fracture is to analyse the minimum principal strain. This criterion is used to predict failure in the FE analysis. As it was proven in a study by Eswaran et al. the correlation of the values of biomechanical tests and the results of the FE analysis. [11][12]

According Eswaran et al. and Mabe et al. the reference value for the vertebral failure is \(-450 \mu \text{m} \). Thus, assuming a failure criterion of minimum principal strain, this is the threshold value used (table 1).

<table>
<thead>
<tr>
<th>Failure criterion for the vertebra</th>
<th>Minimum Principal Strain</th>
</tr>
</thead>
<tbody>
<tr>
<td>Strain</td>
<td>(-4.5 \times 10^{-4} \text{ strain})</td>
</tr>
</tbody>
</table>
3. Methods

In this work the main goal is to define a failure criterion for the vertebral fractures and try to predict the probability of that happen. For this purpose we need to develop a finite element model, so we use CT scan imagens of a human healthy vertebrae to develop the geometric model.

In Figure 2 is presented the methodology followed to obtain the model.

Figure 2 - Methodology used to develop the model.

Since the acquisition of the images there are five general ideas that need to be followed:

1. Image resampling (Mimics),
2. Image segmentation of the resampling (Mimics),
3. Mesh surface adjustments (3-matic),
4. Establishment of solid model (SolidWorks),
5. Generating a volumetric finite element model (Abaqus).

After the creation of the geometric model the file was imported to Abaqus where the finite model is created. For the mesh convergence study, we use a model where the vertebra has two types of material, cortical bone and trabecular bone. In Table 2 it is possible to see the properties of materials in more detail and the element type used for each bone.

<table>
<thead>
<tr>
<th>Material Type</th>
<th>Young's Modulus (MPa)</th>
<th>Poisson's Ratio</th>
<th>Element Type</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical Bone</td>
<td>12000</td>
<td>0.3</td>
<td>M3D3</td>
</tr>
<tr>
<td>Trabecular Bone</td>
<td>100</td>
<td>0.2</td>
<td>C3D4</td>
</tr>
</tbody>
</table>

Table 2 - Material properties and element types.[13][14][15][16][17]

For the assembly of the properties was necessary to define two sections for the differentiation between cortical bone and trabecular bone. In the geometric model we created a shell, an outer surface with a given thickness, that simulated the cortical bone.

Two Constraints were defined, then was possible in the section Load of the software Abaqus add the loading and boundary conditions. The load was set as a concentrated force with a value of 18kN, with vertical direction and as a compressive force. The boundary conditions were defined as Encastre, so the lower surface of the vertebra was fully fixed, allowing no degree of freedom in these points.

For the finite element analysis after we define the best mesh to use through a suitable convergence study, we used a bone remodelling model to define the material properties. In this case the material properties are anisotropic, heterogenic and viscoelastic.

In the model of bone remodelling the shell was removed, but it continued to be distinction between cortical and trabecular bone these distinction was given by the bone remodelling bone. In Table 3 are the material properties that we gave to the bone remodelling model.

<table>
<thead>
<tr>
<th>Material Type</th>
<th>Young’s Module</th>
<th>Poisson Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical Bone</td>
<td>20 GPa</td>
<td>0.3</td>
</tr>
<tr>
<td>Trabecular Bone</td>
<td>100</td>
<td>0.2</td>
</tr>
</tbody>
</table>

Table 3 - Material properties for the FE model of the bone remodelling.

For the bone remodelling the loading conditions were changed, to make the model more complete and to obtain the best results. Then were created several steps in Abaqus model in which the various steps represent the different movements made by the human vertebra.

<table>
<thead>
<tr>
<th>Body Position</th>
<th>Compressive Force (N)</th>
<th>Moment (Nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flexion</td>
<td>1175</td>
<td>7.5</td>
</tr>
<tr>
<td>Extension</td>
<td>500</td>
<td>7.5</td>
</tr>
<tr>
<td>Lateral bending</td>
<td>700</td>
<td>7.8</td>
</tr>
<tr>
<td>Axial rotation</td>
<td>720</td>
<td>5.5</td>
</tr>
</tbody>
</table>

Table 4 - Loading modes for the simulation of different position of the human body. [15]
In Table 4, the values used for each movement are shown, these values were taken from a study by Dreischarf et al., and those values were measured from human experiments.

In the bone remodelling model, we can choose different parameters where the most important for this work is the \( k \) and \( m \) parameters. These two parameters are biological ones that we can change to set the best values that represent the best combination to obtain the most realistic model for a healthy vertebra. \( k \) is the biological cost of the organism in maintaining bone homeostasis and \( m \) is a corrective factor for the preservation of the intermediate densities.

Different simulations with different values for \( k \) and \( m \) were performed until we find the best combinations, the decisions for the other parameters of the bone remodelling model were made based on previous studies.

4. Results and Discussion

4.1 Mesh Convergence Study

For the mesh control (mesh refinement) was used the seeds size. Seeds are points placed along the edges of the model, the distance between them is what we can control in the formation of the mesh. After we choosing this value the mesh was built freely by the software Abaqus.

In the figure 3 two meshes are showed, (A) with fewer elements compared to (B). The red dot is the spot chosen for the convergence analysis. The dot lies in a border region of different tensions, which demonstrates the importance of having a high number of elements because the smaller the area of the closest element closest is the tension value to the reality.

![Figure 3 - Results of two simulations with different meshes, and where we can see the von Mises tension, in (A) for a mesh with the global seed size of 4 and in (B) 1.6. the red point is the point chosen for the convergence analysis.](image)

![Figure 4 - Convergence curve for vertebra L1 and respective logarithmic trend line.](image)

In Figure 4 are showed the chart of the evolution of the tension in the control point with the increase of elements. As we can see the tension value stabilize for a number of elements equal to 58131 and the variation is not so significant after that.

So, it was chosen as the reference value for the global seed size 1.6, with this value we have a mesh with 102762 elements. We could have chosen a smaller value for the seed size but we would have a very high value of elements which despite having better results would put in jeopardy the timeline delivery of the work. A high computing power would be needed, which would increase the time of the analysis we wanted to do in future, so we opted for this mesh by the feasibility demonstrated.

![Figure 5 - Menus of interest of bone remodelling model.](image)
4.2 Bone Remodelling Model Results

In this section we showed the results we find to the best values of \( k \) and \( m \), the values that simulate in a more realistic way the vertebra densities as we can see in a CT scan and we can compare also with other FE models.

The parameters that we define for \( k \) and \( m \) are 0.0006 and 2, respectively. As we can see in the above image the values for the trabecular bone, on the inside of the vertebra at the central area, are between 0.2-0.5 like in the FE models found in literature. In the outermost region, cortical area, was registered higher densities as expected, it is possible to see also that the density distribution increases in the interior of the vertebral body to the outside.

The interior of the vertebra is not uniform in the bone remodelling model like in other models in literature, what is more important to the type of material used to define the properties of the same, rather a material with isotropic properties in which the vertebra is defined homogeneously with only a distinction between trabecular and cortical bone. Thus we see, how the closest to the reality is the use of this model, start with a vertebra with a variable density, where we have the distinction between the cortical part, with higher density and trabecular part, with lower density. And in the final we have a vertebra in which we see that although there is some pattern the inner part does not have a homogeneous density, which translates a real and healthy vertebra.

For understand if the number of the iterations set for the simulation is the best one the model gives to the user a chart with the evolution of the objective function with the iterations. As we can see in Figure 7, after the 50th iteration the model stabilizes, and continue to run the program for more iterations would give a better solution but the computational cost does not improve the results in order to compensate.

![Figure 6 - Results for the distribution of relative density for the bone remodelling bone for a computational simulation for \( k = 0.0006 \) and \( m = 2 \).](image)

Above we show the results for the qualitative analysis, for the quantitative we found in literature a reference value to the bone mass of the vertebra.

For this we analysis the volume of each element and the relative density of them, those values are given by the bone remodelling model. This values are used to calculate the total mass of the vertebra.

The following relations are used to calculated the mass:

\[
\rho_{\text{total}} = \rho_{\text{relative}} \times \rho_{\text{average}}, \quad \text{with} \quad \rho_{\text{average}} = 1.75 \text{g/cm}^3 \tag{5}
\]

\[
\frac{\text{mass}}{\text{Volume}} = \rho_{\text{total}} \tag{6}
\]

<table>
<thead>
<tr>
<th>Total Volume (cm(^3))</th>
<th>Total Mass (g)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Vertebra L1</td>
<td>40.39</td>
</tr>
<tr>
<td>Vertebra L3 (Duan et al.)</td>
<td>52.51</td>
</tr>
</tbody>
</table>

Table 5 - Values for total volume and mass of the vertebra L1 for the simulation of the bone remodelling model with \( k = 0.0006 \) and \( m = 2 \).

In the above table are the results for the healthy case of our work, vertebra L1, and the results of the work of Duan et al. for an average human vertebra L3. [18]
The results are within the expected, our values must be lower than the study of Duan et al. because the L3 is bigger than L1. The value of the volume and the mass that we obtained are smaller but they are pretty close to the reference values, which were obtained from real vertebrae. We have qualitative and quantitative confirmation that our vertebra is very close to a real and healthy vertebra.

4.3 Simulation of Osteoporotic Models
To simulate the osteoporotic models, we rely on a study done by Santos et al. that defined a relationship between the value of the \( k \) and \( T \)-score, this value is obtained from the BMD obtained through a densitometry (DXA). This work started from a database with different tests of DXA and through the simulations with the bone remodelling model we used, obtained a correlation between \( k \) and BMD. Through the database and the results from the simulation was found a relationship between the value \( k \) and \( T \)-score[19]:

\[
T - score = 55.931k^2 - 35.584k + 2.334 \quad (7)
\]

With this relationship was possible obtain a relation between the \( T \)-score value and \( k \), for the healthy and osteoporotic cases.

\[
k_{\text{osteoporotic}}^\frac{k_{\text{healthy}}}{k_{\text{healthy}}} = 1.7 \quad (8)
\]

\[
k_{\text{osteoporotic, maximum}}^\frac{k_{\text{healthy}}}{k_{\text{healthy}}} = 3 \quad (9)
\]

The healthy \( T \)-score is -1.1, for values equal or higher than -2.5, in modula, is already considered osteoporotic, in this study the maximum value considered to \( T \)-score was -3.3.

With this relationship we set the \( k \) value for the osteoporotic case and for the more severe case, so with these values we made 5 simulations to pathological values, increasing the severity of cases, for that we increase the value of \( k \), by increasing this value we decreased the bone density, which simulates in a simplified way osteoporosis.

Table 6 - Values of parameter \( k \) used for the simulations of bone remodelling model for pathological cases.

<table>
<thead>
<tr>
<th>( k ) value</th>
<th>Osteopenia</th>
<th>Osteoporosis</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0.00081</td>
<td>0.00102</td>
</tr>
<tr>
<td></td>
<td>0.00128</td>
<td>0.00154</td>
</tr>
<tr>
<td></td>
<td>0.0018</td>
<td></td>
</tr>
</tbody>
</table>

Figure 8 - Results for the simulations of different values of \( k \) of the relative density, in (A) the healthy case \((k=0.0006)\), in (B)(0.00102), (C)(0.00154) and (D)(0.0018) three osteoporotic cases.

In Figure 8 we can see the results of our models, we can compare the healthy case with three osteoporotic cases, in our case it is not possible compare the results at the microstructure level, but we can see the areas and the values of relative density.

As we can see there is a gradual decrease of the relative density inside the vertebra, trabecular region, in cortical level we can also see that this area is getting thinner, and from (A) to (D) the difference is significant, beyond getting thinner the value of relative density decreases.

As expected the trabecular structure weakens, we can associate the blue regions (lower relative density) with areas in which the vertebra loses the trabeculae or where the trabeculae are thinner and longer.
4.3.1 Failure Criterion: Minimum Principal Strain

In the following images are showed the results for different steps for two cases, the healthy one and an osteoporotic case. The results showed the minimum principal strain that is the failure criteria defined for the vertebra.

![Figure 9 - Results of the minimum principal strain for the healthy vertebra, for different steps.](image1)

![Figure 10 - Results of the minimum principal strain for the most severe case of osteoporotic for different steps.](image2)

In the osteoporotic case it is possible to see the regions where the failure criteria are exceeded, $\epsilon_{\text{min}} = -4.5 \times 10^{-4}\text{strain}$. These regions are the most susceptible to the occurrence of a fracture. Comparing with CT scan imagens of fracture vertebrae in the majority of the cases the fracture stars in the anterior part of the vertebral body and depending on the severity the fracture can spread to the rest of the vertebral body and this could collapse completely.
The study of osteoporosis and vertebral fracture are of major interest, due to their high prevalence in population, the decrease in life quality of the patient and their social and economic impact. Combining the FE models and bone remodelling bone allows a better comprehension of the disease and the solution for this problem, so we present some conclusions and future directions:

- FE models can be very important in clinical context and help physicians to intervene earlier in these cases, that is, before fractures occurs, because the treatment is complicated and the recovery is very slow. If the intervention occurs before the fracture the method is easier, the patient recover earlier and gain life quality.

- In the FE models the larger the area with higher minimum principal strain values (in module) to the value of the failure criterion for the vertebral fracture more likely of fracture.

- With this model, a door opens to attempt predicting the occurrence of vertebral fractures due to the osteoporotic level of the patient, an earlier intervention making it possible, increasing the chances of a faster recovery and an improving life quality.

- In the future experimental work can be done, biomechanical tests, with human vertebrae to the validation of the FE models.

- In our model we just used a vertebra to developed the model, the next step could be developed a FE model with a functional spinal unit (FSU), the smallest physiological motion unit. FSU consists of two adjacent vertebrae, the intervertebral discs and all the ligaments between them. Or use other region of interest or maybe all the spine.

- Simulations with healthy, osteoporotic and intervened surgically vertebrae, make appropriate comparisons and realize if the methods used to the prevention/treatment of osteoporotic vertebral bodies or the fractured ones are the most appropriate and whether that need changes or not.

- Data base with vertebrae CT-scan and bone densitometries with the BMD and T-score associated, for comparison between the model obtained in the FE model created through CT-scan images and bone remodelling model, and this comparison it was a validation method.

- Developing a model that could be used in clinical practice, the physician introducing the patient's CT-scan, the geometric model was created and the bone remodelling analysis was made. In the end through a defined failure criterion and a defined values range the physician had the most susceptible areas suffer a fracture in the future. A decisions support model to help the physician in clinical practice. This could rise the success of the intervention by given the physician a second opinion to support his decision.

References


Based on the Trajectorial Theory of Wolff.,”


[18] Y. Duan, a M. Parfitt, and E. Seeman, “Vertebral bone mass, size, and volumetric density in women with spinal fractures.,”