The osteoporotic vertebral structure is well adapted to the loads of daily life, but not to infrequent “error” loads

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Abstract

Osteoporotic vertebral fractures typically have a gradual onset, frequently remain clinically undetected, and do not seem to be related to traumatic events. The osteoporotic vertebrae may therefore be expected to display a less “optimal” bone architecture, leading to an uneven load distribution over the bone material. We evaluated the trabecular load distribution in an osteoporotic and a healthy vertebra under normal daily loading by combining three recent innovations: high resolution computed tomography (µCT) of entire bones, microfinite element analyses (µFEA), and parallel supercomputers. Much to our surprise, the number of highly loaded trabeculae was not higher in the osteoporotic vertebra than in the healthy one under normal daily loads (8% and 9%, respectively). The osteoporotic trabeculae were more oriented in the longitudinal direction, compensating for effects of bone loss and ensuring adequate stiffness for normal daily loading. The increased orientation did, however, make the osteoporotic structure less resistant against collateral “error” loads. In this case, the number of overloaded trabeculae in the osteoporotic vertebra was higher than in the healthy one (13% and 4%, respectively). These results strengthen the paradigm of a strong relationship between bone morphology and external loads applied during normal daily life. They also indicate that vertebral fractures result from actions like forward flexion or lifting, loads that may not be “daily” but are normally not traumatic either. If future clinical imaging techniques would enable such high-resolution images to be obtained in vivo, the combination of µCT and µFEA would produce a powerful tool to diagnose osteoporosis.

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Introduction

Osteoporosis is the most common bone disease and its clinical significance lies in the high susceptibility to fracture. Although osteoporotic fractures can occur anywhere in the human body [18], one of the sites with the highest prevalence is the spine, particularly in the elderly population [19]. In Europe, the incidence of new vertebral fractures among people aged 50–79 years amounts to 1.1% in women and 0.6% in men [5]. From a mechanical point of view, these vertebral fractures must be caused either by reduced structural strength of the vertebral bodies or by increased loads. As most vertebral fractures have a gradual onset and remain clinically undetected, they are thought to be the result of normal daily loads rather than traumatic events [19]. There are several possible explanations for such a non-traumatic fracture. First, because the osteoporotic vertebrae display a reduced bone mass, there is less bone for load distribution, resulting in higher local tissue stresses and an increased risk of fracture. A second explanation is that osteoporotic vertebral trabecular bone exhibits an altered trabecular micro-architecture with less transverse trabeculae [14,20]. This altered architecture could result in an uneven distribution of the load within the vertebrae, with some trabeculae...
virtually unloaded and others overloaded. Third, the tissue strength of osteoporotic bone may be reduced, increasing the risk of fracture even for normal tissue loads. In a previous study, we found that a change in the average tissue yield strength due to osteoporosis is unlikely [12], but the mechanism of vertebral fracture is still unknown. For this study, we asked the question whether vertebral fractures are due to typical, common loads encountered daily (the first suggested cause) or to typical “error” loads such as those resulting from a fall or stumble (the second suggested cause).

For that purpose, we determined stresses and strains in the trabecular structures of one normal and one osteoporotic vertebra, and compared those for both normal (daily) and “error” loads. To accomplish that, we used a combination of three recent innovations for the evaluation of trabecular stress distributions throughout the entire human vertebral bodies. The first one is microcomputed tomography (μCT) scanning, which permits digitization of entire human bones three-dimensionally at resolutions of around 60 μm. This technique allows one to image the thin cortical shell as well as all trabeculae within the vertebral body. The second innovation is that of microfinite element analyses (μFEA), where standard (commercial) FEA codes can conveniently handle up to 10^5 elements. These large-scale FEA codes permit computations for up to 10^9 elements. With these new codes, analyses of entire human bones at high resolutions have come into reach, albeit computer time is still excessive [32]. The third innovation is the use of new supercomputers with large quantities of parallel processors. By distributing the analyses over multiple processors, the calculations can be accelerated substantially. The combination of these three developments offers a novel and very powerful analytic modality for bone biomechanics.

### Methods

In previous studies by Bürklein et al. [2] and Lochmüller et al. [15], a large set of more than 100 vertebral bodies was obtained from elderly donors. In these studies, the correlation between DXA and QCT measurements and structural strength was determined for both thoracic and lumbar vertebrae. From this set of specimens, two lumbar vertebrae (L1) were selected based on the compression strength of the L3 vertebra and bone mineral density (BMD) status of the segment L2–4. The osteoporotic vertebra originated from an 81-year-old female (weight: 53 kg, height: 1.65 m) with an L3 compression strength of 2208 N and a T score of −4.4 (WHO criterion). The healthy vertebra came from an 80-year-old female (weight: 62 kg, height: 1.58 m) with an L3 compression strength of 4157 N and a T score of +2.5. Table 1 lists details of these two individuals, specifically age, weight, and height as well as the T scores of the lumbar vertebral segments L2–4 (from in situ AP DXA) and structural strength (failure load) of T6, T10, and L3. Further details about the vertebrae can be found in the papers mentioned above [2,15].

In the current study, both vertebrae were scanned with a μCT scanner (μCT80, SCANCO Medical AG, Switzerland). The resulting three-dimensional images were thresholded using an individual value that gave the best visual agreement with the raw μCT-images. Voxel finite element models [31] with 33 million (model of healthy case) and 25 million (model of osteoporotic case) cubic brick elements of 65 μm were obtained from these three-dimensional images (Fig. 1). The upper and lower endplates were augmented with soft layers that were solely intended to properly distribute the load over the endplates. Through these soft layers a total load of 1.2 times bodyweight (peak load during walking [25,35]) was distributed such that the stresses through the nucleus region of the disc were 30% higher than those through the annulus region [1,33]. All bone element material properties were assumed isotropic and linear elastic, with a tissue...
stiffness of 18 GPa \[23\] and a Poisson’s ratio of 0.3. Following the work of Mosekilde \[21,22\], we specified both the outer shell and the spongiosa as made out of trabecular bone material. The finite-element models were solved using a specific microfinite element code \[31\], 16 parallel processors on a supercomputer (SGI Origin 3800, Silicon Graphics, USA), and about 8 weeks wall clock computing time (about 20 000 h CPU time). In both vertebrae, we determined the strain and strain energy density (SED) distributions throughout the vertebral bodies for all individual trabeculae. Specifically, we determined the contribution of the spongiosa to the total load. We obtained three-dimensional plots of the strain and SED distributions, and strain and SED frequency plots.

Subsequently cancellous bone cubes (\(\pm 10 \times 10 \times 10\) mm) were selected from the center of both vertebrae to determine common microstructural parameters: volume fraction (BV/TV), trabecular number (Tb.N.), trabecular thickness (Tb.Th.), trabecular spacing (Tb.Sp.), structure model index (SMI), mean intercept length in anterior–posterior (MILAP), medial–lateral (MILML), and superior–inferior (MILSI) direction, and the degree of architectural anisotropy (DA).

Results

All architectural parameters were different in the osteoporotic trabecular structure as compared to the healthy structure (Table 2). The largest relative differences were found for the volume fraction (\(-31%\) in the osteoporotic case) and the degree of architectural anisotropy (+24%). Smaller differences were found for trabecular number (\(-13%\)), trabecular thickness (\(-11%\)), trabecular spacing (+16%), and the mean intercept lengths (AP: +3%, ML: −3%, SI: +12%). Compared to data from the BIOMED I project, the differences in volume fraction, trabecular thickness, trabecular spacing, and degree of anisotropy were larger than one standard deviation \[28\].

The contribution of the spongiosa to the total load depended strongly on location within the vertebrae. Near the endplates, the spongiosa carried most of the load, whereas toward the middle of the vertebrae the load was more evenly distributed between the shell and the spongiosa. In the healthy case, the contribution of the trabecular core was larger than in the osteoporotic case (Fig. 2). Larger principal strain values (\(<−750\) \(\mu\)strain or \(>750\) \(\mu\)strain) were seen more commonly in the vertical trabeculae. The low

![Fig. 2](image-url)

Fig. 2. The load through the spongiosa as a function of height throughout the vertebral body. The solid line represents the healthy vertebra and the dashed line represents the osteoporotic vertebra.

<table>
<thead>
<tr>
<th></th>
<th>BV/TV (%)</th>
<th>Tb.N. (−)</th>
<th>Tb.Th. (µm)</th>
<th>Tb.Sp. (µm)</th>
<th>SMI (−)</th>
<th>MILAP (−)</th>
<th>MILML (−)</th>
<th>MILSI (−)</th>
<th>DA (−)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Healthy</td>
<td>14</td>
<td>0.99</td>
<td>268</td>
<td>957</td>
<td>2.51</td>
<td>0.31</td>
<td>0.33</td>
<td>0.36</td>
<td>1.15</td>
</tr>
<tr>
<td>Osteoporotic</td>
<td>10</td>
<td>0.87</td>
<td>238</td>
<td>1111</td>
<td>2.65</td>
<td>0.30</td>
<td>0.29</td>
<td>0.41</td>
<td>1.43</td>
</tr>
<tr>
<td>Relative difference</td>
<td>−31%</td>
<td>−13%</td>
<td>−11%</td>
<td>+16%</td>
<td>+5%</td>
<td>−3%</td>
<td>−12%</td>
<td>+14%</td>
<td>+24%</td>
</tr>
</tbody>
</table>

principal strains (−50 to 50 μstrain), on the other hand, were located in both vertical and horizontal trabeculae alike (Fig. 3). The strain and SED frequency plots differed very little between the two vertebrae (Fig. 4). In the healthy vertebra, 91% of the tissue encountered strain levels between −400 and +400 μstrain. Interestingly, this was not different in the osteoporotic vertebra (92%) (Fig. 4, shaded area). In both the healthy and the osteoporotic vertebra, 72% of the material...
experienced strain energy density values below \(0.5 \times 10^{-3}\) MPa (Fig. 5, shaded area).

For the “error”-loading modes, the osteoporotic trabecular bone displayed more highly loaded trabeculae than the healthy cancellous bone. While 4% of the tissue encountered strains smaller than \(-400\) or larger than \(+400\) \(\mu\)strain in the healthy case, as much as 13% of the bone tissue was “overloaded” in the osteoporotic case (Fig. 6, unshaded area). In the healthy case, 17% of the material displayed SED values above \(0.5 \times 10^{-3}\) MPa, while for the osteoporotic cancellous bone this value amounted to 29% (Fig. 7, unshaded area). The osteoporotic vertebrae had more than three times as much tissue with SED values above \(2.0 \times 10^{-3}\) MPa than the healthy one (Fig. 7, ultimate right column).

Discussion

Here we present the first finite-element study evaluating trabecular loading in entire vertebral bodies. Much to our surprise, the number of highly loaded trabeculae was not higher in the osteoporotic vertebra than in the healthy one for a normal daily loading situation. Given that, compared to the healthy vertebra, the osteoporotic vertebra had 25% less bone material to carry the load with, this represents an intriguing finding. We expected that the increased fracture risk of osteoporotic vertebrae could be explained by a suboptimal architecture that would result in lower stresses in some parts but higher ones in other parts of the bone when analyzing common axial loading configurations. However, no differences were found; taking body weights into account, this suggests that both are equally well adapted to the normal loads of daily life. The architectural parameters show how the osteoporotic cancellous bone has accomplished this in spite of its lower volume fraction. Compared to the architecture of the healthy vertebra, the trabeculae were fewer in number, thinner, further apart, but more oriented in the axial direction, at the cost of the transverse directions. The effect of this can be seen in Figs. 5 and 6, where the osteoporotic vertebra displays less bone with zero strain or SED values, but more material with low strain and SED values. This was due to the increased orientation in the osteoporotic bone and compensates for the effects of reduced BMD or volume fraction. Such a mutually compensating effect of volume fraction and architecture has also been observed in cancellous bone adaptation during growth [27]. Seen in that light, the osteoporotic bone appears to be equally adapted to common daily loads as healthy bone and the effects of normal daily loads cannot explain the increased fracture risk of osteoporotic vertebrae.

Fig. 5. Tissue strain energy density frequency plots of the entire vertebrae under the peak load during walking. The black line represents the healthy vertebra and the gray line the osteoporotic vertebra. Note how the osteoporotic vertebra displays less material with zero strain values, but more material with low strain values (the lower bars at zero and the higher bars just next to zero).

![Fig. 5](image1)

Fig. 6. Tissue principal strain frequency plots of the trabecular bone subvolumes under an “error” load resulting from forward flexion and lifting. The black line represents the healthy bone and the gray line the osteoporotic bone.

![Fig. 6](image2)
However, longer and thinner trabeculae are more vulnerable to buckling, thinner trabeculae are more likely to be fully perforated by osteoclasts [30], and the axial “over-orientation” makes the structure particularly vulnerable to loads in unusual directions. As the increased vulnerability to buckling becomes particularly important for high loads, this may explain the decreased compression strength that was measured for the osteoporotic L3 vertebra. This also implies that fractures of osteoporotic vertebrae may be caused by loads that are higher than normal. Vertebral fractures may on the other hand also be caused by infrequent loads in unusual directions, “error loads”, which are not trauma related.

The observed changes in the architecture likely lead to an increased vulnerability for such “error” loads. This was confirmed when simulating an “error” load, as occurs in forward flexion or lifting. The results of these analyses showed a less optimal stress transfer in the osteoporotic bone only for uncommon error loads, but not for typical axial loads.

The contribution of the spongiosa in load transfer depended strongly on the location within the vertebra. Near the endplates, the spongiosa carried most of the load, but toward the middle of the vertebra the load became more evenly distributed over the spongiosa and the shell. The main reason for this lies in the architecture of the shell and the trabeculae within the spongiosa. Going from the endplates toward the middle of the vertebra, the shell curves inward, while simultaneously the outer trabeculae angle from the endplate towards the shell. As a result, the mid-transverse shell “collects” the load from the outer trabeculae. A similar load shift was observed in previous computational studies [11,26] and was confirmed here for whole vertebral bodies. Although the contribution to load support of the spongiosa was larger in the healthy vertebra than in the osteoporotic one, both fell within the range of values found previously for both healthy and osteoporotic vertebrae [6,11,16,24,26,36].

This study involved several limitations: first, we studied only one healthy and one osteoporotic vertebra; our results are thus exemplary. Although we expect the computational expenses for large-scale µFEA studies on whole bones to decrease dramatically within a few years, they presently defy inclusion of larger series. However, as the two vertebrae represent extremes concerning T scores (−4.4 and +2.5) and were carefully selected from a large sample, we expect them to be representative examples. Second, in our finite-element calculations, we assumed the tissue material to be homogeneous and equal for both vertebrae. We have previously demonstrated that trabecular tissue stiffness and strength are on the average unchanged by osteoporosis [12]. Recently, McCreadie [17] has, however, shown that tissue mineralization is inhomogeneous and that the inhomogeneity is higher in osteoporotic patients. Assuming that this leads to higher variation in tissue elastic properties, the tissue level stresses and strains will show more variance than calculated with our models. This will, however, only marginally influence the frequency plots [7,31]. The results of previous studies [13,17,29] suggest that the effects of inhomogeneous mineralization on the strength of the vertebra are likely very small. However, it cannot be ruled out that if the inhomogeneity of tissue strength is increased as well, local failure may occur sooner, which may change the strength of the vertebra as a whole. Third, our linear finite-element method does not allow for the analysis of trabecular buckling. Although buckling is not likely to play a role for the normal daily loads that were investigated here, it may be important when investigating the effects of higher loads. For this reason, we were not able to computationally determine the strength of the vertebrae as found in the compression tests. Fourth, the boundary conditions that were applied to the cancellous bone cubes only take into account the weight difference between the donors. Possible differences in the load-carrying behavior of the shell and endplates were excluded from these calculations simply because they are unknown for “error” loads.

In conclusion, our finite-element models indicated that the osteoporotic vertebrae are equally well adapted as the healthy ones for normal daily loads, but run a higher risk of failure when subjected to uncommon “error” loads. These results strengthen the paradigm of a lasting, strong relationship between bone morphology and external loads throughout life. They also suggest that vertebral fractures may originate from actions like forward flexion or lifting, loads that may not be “daily” but are also not normally traumatic.
Acknowledgments

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