On the uniformity of stresses and strains in the femoral head

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According to Wolff's trajectorial hypothesis[(1)], the trabecular architecture is such that minimal tissue stresses are paired with minimal weight. This paradigm at least suggests that, normally, stresses and strains should be distributed rather evenly over the trabecular architecture. Until recently, however, there have been no possibilities for a quantitative evaluation of this paradigm. It has been argued that the premise of Wolff's law is a false one and that the correspondence between trabecular architecture and stress trajectories calculated from the 'Graphical Statics'[(2)], on which it is based, is just an optical illusion[(3)]. The 'Graphical Statics' technique for analysis of stress transfer in solids is based on requirements for continuity of the material. The results of these analyses only tell us about the courses of the stress trajectories if the bone would be made out of a single homogeneous and isotropic material with no trabecular architecture. It has been argued as well that the similarity between trabecular orientation and stress trajectories is circumstantial rather than causal and that there are no mathematical rules for bone architecture[(4)]. Over the last decades, far more advanced computational models, as based on finite element (FE) analysis, were developed to calculate stresses and strains in complex three-dimensional structures. Where analyses of bones are concerned, however, this approach still suffers from the same limitations: bone material can only be represented as a homogenized continuum. Although such methods can account for the local porosity and anisotropy of the material, though they can only provide stresses and strains at the homogenized level, and not those in individual trabeculae.

Recently, however, new numerical approaches have made it possible to calculate the strains at the level of the trabeculae itself[(5)]. With this technique, the architecture of the trabecular bone is quantified by a large number of sequential cross-sectional images. These digitized images are stacked in a computer in which the 3-D trabecular structure is reconstructed as a rectangular voxel grid, with voxels representing bone tissue or marrow. By converting voxels representing bone tissue to equally shaped brick elements in a micro- finite element model, a model is generated that can represent the trabecular structure in great detail (Fig. 1). When realistic forces are applied to the model, the micro-FE analyses provide stresses and strains at the level of the bone tissue. This technique has enabled us to investigate the paradigm that bone tissue-level stresses and strains should be evenly distributed. In this short note, I will summarize the results of two studies in which these

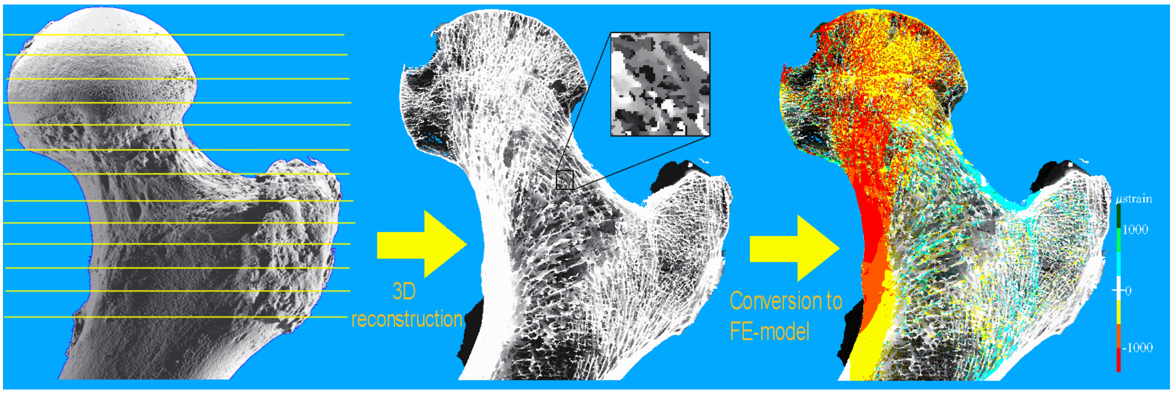

Fig. 1Overview of the micro- FE approach strains were investigated for a human and canine femoral head and note some remarkable similarities in these results.

In the first of these studies, the stresses and strains at the level of the trabeculae were calculated for a canine femoral head[6]. In this study, micro-CT images were used to create a micro- finite element (micro-FE) model of the proximal part of the femur that can represent the 3-D trabecular architecture in great detail. Micro-FE analyses were conducted for three different orthogonal hip-joint loading cases, one of which represented the stance-phase of walking. By superimposing the results, the tissue stress and strain distributions could also be calculated for other force directions. For the stance phase of walking an average tissue principal strain in the VOI of 279 microstrain was found, with a standard deviation of 212 microstrain. The standard deviation depended not only on the hip-force magnitude, but also on its direction. No single load creates even stress or strain distributions in the trabecular architecture.

In the second study, tissue-level stresses and strains were calculated for trabecular bone in a human femoral head[7,8]. As in the previous study, a micro-FE model of a healthy proximal femur was made from micro- computed tomography images that could represent the internal architecture of the bone in detail, resulting in a model with over 96 million elements. The model was loaded by a distributed force acting on the femoral head that represented the hip-joint force during the stance-phase of walking. The average tissue-level principal strain magnitude in the femoral head was 304 (S.D. 185) microstrain.

When comparing the average magnitudes of the largest principal strain component, the largest principal stress component, the Von Mises stress, and the strain energy density (SED), for the femur studies, remarkable similarities are found (Table 1). The average strain magnitudes found for the canine and human femoral head differ by only 8%. Differences between the average Von Mises stress and SED value are only about 1%. In neither of these studies, however, a perfectly even distribution was found for the parameters investigated. This suggests that the bone architecture is likely adapted to more diverse loading rather than to a single load case. This could also explain why the variation in tissue-level stresses and strains in the canine femur was larger than in the human femur; hip-joint force in the canine vary considerably in direction and magnitude during the walking cycle, whereas that in the human varies in a much more limited range[9,10].

Some limitations of these studies need to be discussed. First, the values of the bone tissue-level stresses and strains found in these micro-FE models are linearly related to the hip-joint force applied to the models. The hip joint force in both humans and dogs has been accurately quantified from in-vivo telemetry measurements as a function of body-weight[9,10]. However, these studies have also demonstrated considerable variation in stance-phase hip-joint forces between individuals, up to about 35% in humans[10]. Second, the values are dependent as well on the material properties specified for the bone tissue itself. In both studies, the bone tissue was given the same Young's modulus (15 GPa). In recent studies, higher values have been reported for the bone tissue modulus[11], which could imply that the strains in the studies presented here are overestimated. Third, the results obtained here cannot be translated directly to other sites, which might be adapted to other stress and strain values than the femoral bone investigated here. For example, it has been suggested that tissue-level stresses and strains for bone in the skull should be much less than those in the weight-bearing skeleton[12].

Although available for only two femoral heads so far, these results provide a first indication that stress and strain values at the level of the bone tissue could be rather uniform, even between species. It is expected that these results can be used to formulate objective functions for optimization or remodeling studies that aim at predicting bone architecture.

Table 1: average values (standard deviations) of stresses and strains in the femoral head

	Principal strain magnitude [microstrain]	Principal stress magnitude [MPa]	Von Mises stress [MPa]	SED [MPa]
Canine [6]	279 (212)	3.88 (3.04)	4.60 (3.36)	1020 (1480)
Human [7,8]	304 (185)	*	4.66 (2.80)	1011 (1620)

* not reported

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