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The Influence of Muscle Forces on the Stress Distribution in the Lumbar Spine

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Abstract: *Introduction*: Previous studies of bone stresses in the human lumbar spine have relied on simplified models when modeling the spinal musculature, even though muscle forces are likely major contributors to the stresses in the vertebral bones. Detailed musculoskeletal spine models have recently become available and show good correlation with experimental findings. A combined inverse dynamics and finite element analysis study was conducted in the lumbar spine to investigate the effects of muscle forces on a detailed musculoskeletal finite element model of the 4th lumbar vertebral body.

Materials and Methodology: The muscle forces were computed with a detailed and validated inverse dynamics musculoskeletal spine model in a lifting situation, and were then applied to an orthotropic finite element model of the 4th lumbar vertebra. The results were compared with those from a simplified load case without muscles.

Results: In general the von Mises stress was larger by 30 %, and even higher when looking at the von Mises stress distribution in the superio-anterior and central part of the vertebral body and in the pedicles.

Conclusion: The application of spine muscles to a finite element model showed markedly larger von Mises stress responses in the central and anterior part of the vertebral body, which can be tolerated in the young and healthy spine, but it would increase the risk of compression fractures in the elderly, osteoporotic spine.

Keywords: Lumbar spine, muscle influence, inverse dynamics, finite element analysis.

INTRODUCTION

Precise assessment of fracture risk and risk of other spinal injuries during occupational, athletic and daily activities as well as effective prevention and treatment of spinal fractures depend upon accurate estimation of muscle forces, bone strength, internal and external spinal loads. Such assessments can be acquired by *in vivo* experimental attempts, but remain invasive, costly and often difficult to perform for ethical reasons. Indirect estimation of spinal muscle forces and internal loads has been carried out by measuring intradiscal pressure [1-3] or by using instrumented fixation systems capable of measuring loads [4-6] .

Another approach is computational biomechanical modeling allowing for determination of load contributions from passive and active components of the human trunk and spine in various activities of daily living. However, the influence of muscle forces has been simplified or even neglected in many finite element (FE) models of the spine [1,7-9] or typically represented by a follower load [10,11]. The significance of these simplifications on the simulated bone stresses is still not fully understood, and this is the motivation of the present work.

More precisely, the purpose of this study was to assess the effect of the spinal and trunk muscle forces in regards to the von Mises stress distribution on the 4th lumbar vertebral body in a finite element model (FE L4), thus providing a guideline about the necessisity of modeling muscular complexity of computational FE models, when used for, e.g., fracture assessment, planning of surgical procedures and design of spinal fixation devices. The simulated situation was a standing posture performing a lifting task. The data from the musculoskeletal simulation were then transferred to the FE L4 using a novel interfacing approach to see the effect on the stress distribution.

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MATERIAL AND METHODOLOGY

The Musculoskeletal Model

The standing model of the AnyBody Managed Model Repository version 1.0 (www.anyscript.org) was set up to resemble a 50th percentile European male of 1.75 m and 75 kg. The model comprises roughly 1000 individually activated muscle fascicles of which about 188 are located in the lumbar spine area, thus providing a detailed representation of the distribution of muscle forces attacking the lumbar vertebrae. Detailed information about the lumbar spine model is available from Hansen *et al.,* and Zee *et al.,* [12,13]. The lumbar spine model is validated for a range of activities of daily living by means of comparing computed intradiscal pressures to *in vivo* measurements [14].

Musculoskeletal models must address the so-called redundancy problem [15], stemming from the fact that the muscle system has more muscles than degrees of freedom and therefore is statically indeterminate. In general two commonplace methods are used to overcome this problem; an optimization and an EMG-assisted approach. In the optimization approach it is assumed that the central nervous system will use an optimality criterion to resolve the indeterminacy [16]. In the EMG-driven approach surface electromyography (EMG) signals are measured on a limited number of trunk muscles. Estimates of force-EMG relationships are employed to estimate the spinal muscle forces while trying to satisfy a set of equilibrium equations [17,18]. Also, combinations of the two approaches have been proposed [18-21]. The model of this study used an optimization approach to resolve the indeterminacy problem, in which the cubed sum of muscle stresses was minimized.

This model was modified to represent a situation of daily living as seen in Fig. (**1**), more precisely a lifting scenario with dumbbells of 5 kg in each hand and a posture of 60 degree flexion, 15 degree lateral bending, and 25 degree axial rotation between thorax and pelvis. Muscle forces and joint reaction forces were subsequently computed by the AnyBody Modeling System ver. 4.0 (AnyBody Technology, Aalborg, Denmark) using the methods described by Damsgaard *et al.,* [22].

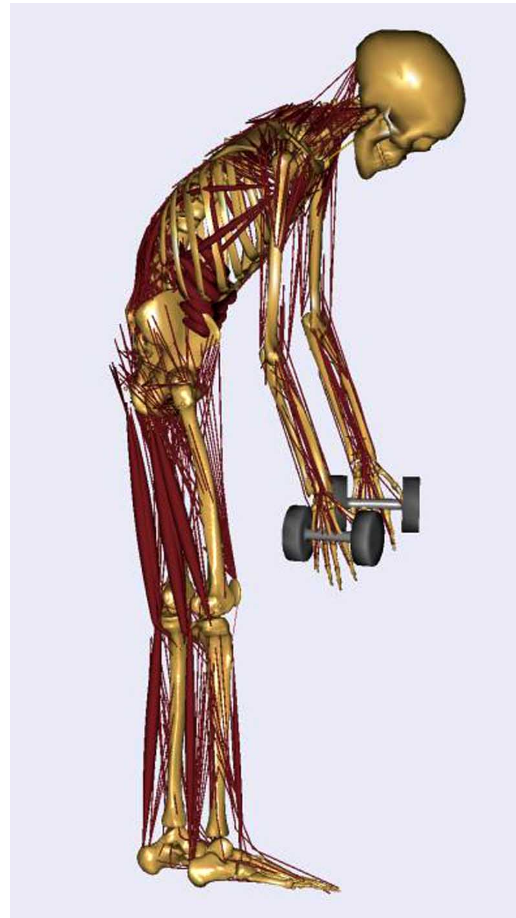


Fig. (1). Inverse Dynamics Model of a lifting scenario with dumbbells of 5 kg in each hand.

The Finite Element Model

FE analysis was performed by the finite element code ANSYS version 12.0®. The 3-dimensional FE model included the 4th lumbar vertebral body. The volumetric part is modelled with 63974 higher order 10-node elements with quadratic displacement behaviour. The volumetric model was generated from a surface geometry derived from computed tomography scans. Shell elements were placed on the surface of the volume to model the cortical shell. Therefore, 8886 8-node shell elements with a thickness of 0.6 mm were used. The material properties for trabecular bone were orthotropic and for cortical bone isotropic. Material properties are shown in Table **1**.

Convergence test for the FE model was performed. This was performed to ensure that the FE model had an appropriate number of elements. When this test is performed, the number of elements in the FE model is increased until a certain point, where the calculated results converge to the exact solution, thus giving the appropriate number of elements to use for the FE model. Please notice that the FE model does not fully comply to state-of-the-art in the sense that it does not have detailed and spatially varying anisothropic material models for the trabecular and cortical parts of the bone. However, for the purpose of this paper, which is merely to assess the effect of excluding versus including the muscle forces, the model is adequate. Fig. (**2**) shows the FE model in a superiolateral view.

Table 1. Material Properties for the Bony Components

Cortical bone	$E = 12000$ N/mm^2 $\upsilon = 0.3$
Cancellous bone	$Ex = 250$ N/mm^2 (ant-post) $Ey = 500$ N/mm^2 (inf-sup) $Ez = 250$ N/mm^2 (med-lat) $\upsilon = 0.3$
Pedicle and vertebral arch	$E = 5000$ N/mm^2 $\upsilon = 0.3$

E = Young's modulus
υ = Poisson's ratio

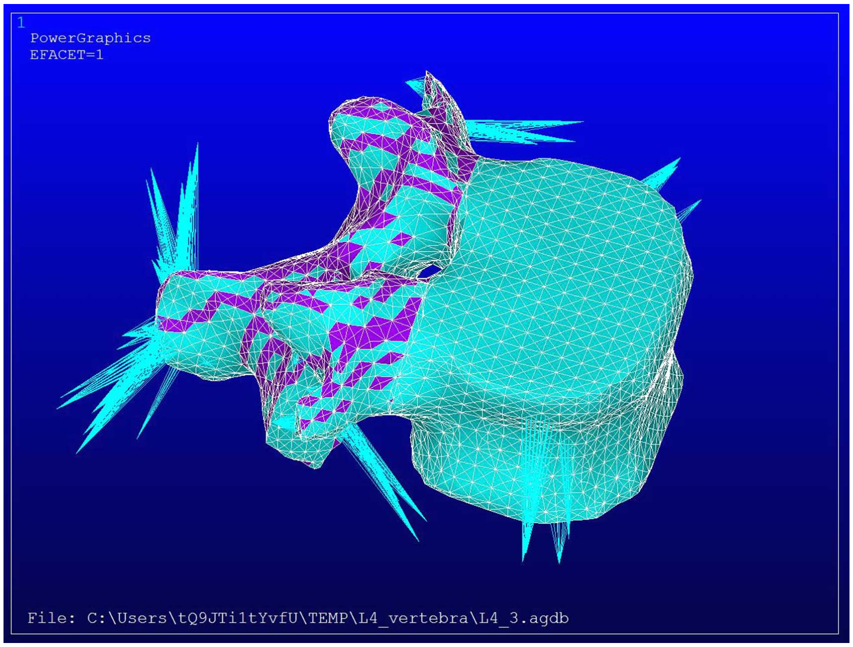


Fig. (2). A FE model in a superiolateral view.

Two separate load cases were applied to the FE model. The first load case had all muscles and joint forces (MUSload). It included a total of 61 individual forces, which were applied to adequate anatomical muscle insertions on the surface of the vertebral body and connected through beam elements with a high stiffness, they can therefore be considered rigid. Fig. (**3**) shows a graphical representation of joint forces and individual muscle forces acting on the L4-5 disc.

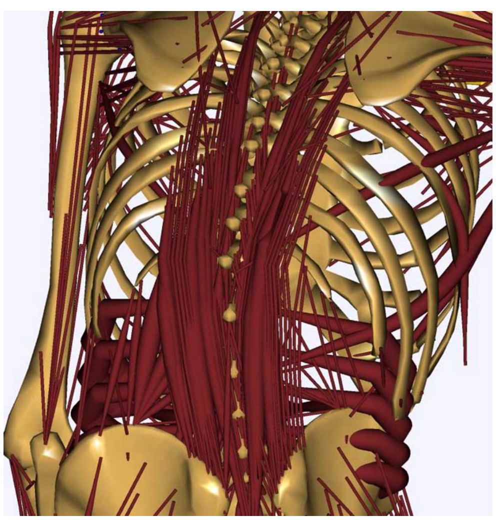


Fig. (3). A graphical representation of joint forces and individual muscle forces acting on the L4-5 disc.

The second load case was a simplified load of only an inferior and superior loading with the same magnitude of joint reaction in case 1 (SIMload). Fig. (**4**) shows the forces acting on the L4-5 disc.

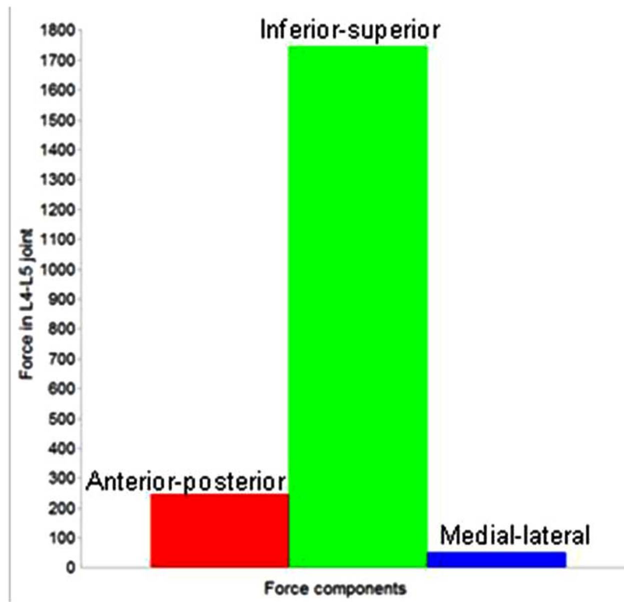


Fig. (4). The forces acting on the L4-5 disc.

RESULTS

The von Mises stress distribution was evaluated by direct comparison of the results of the two load cases. The results are represented in Fig. (**5**) for the SIMload and MUSload load cases.

In general the Von Mises stress distribution was 30 % larger for the MUSload load case compared to the SIMload load case. The peak stresses were markedly larger in the central and upper anterior part of (superior-anterior) the vertebral body. There was also a markedly larger peak stress in the pedicles of the MUSload model. The difference in peak stresses was 2.8-3 MPa for the vertebral body and 5-10 MPa higher in the pedicles.

DISCUSSION

The von Mises stress distribution was markedly different between the two models. In general the MUSload model had

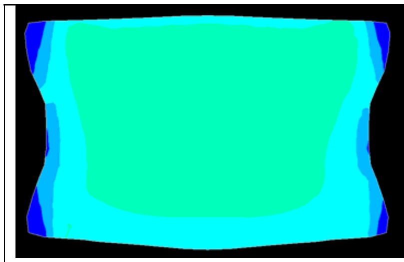
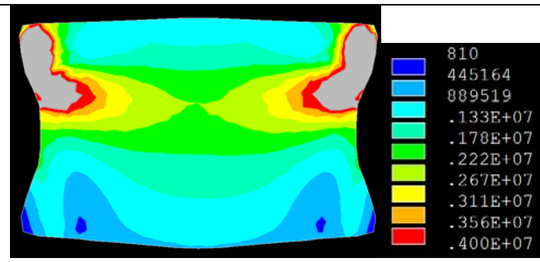
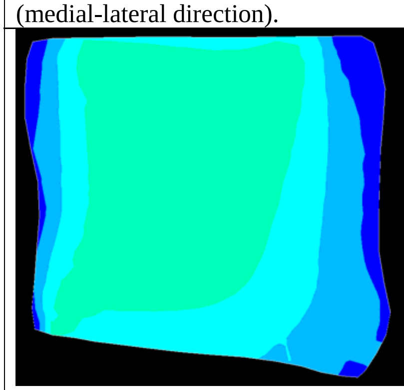
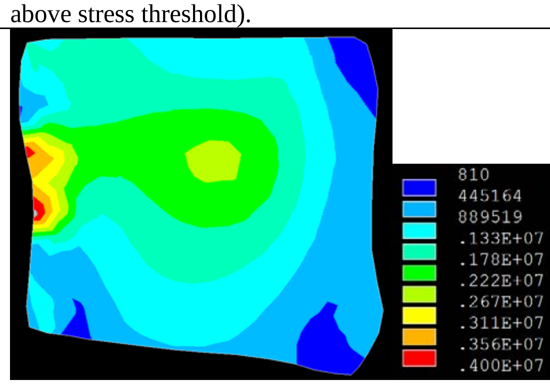
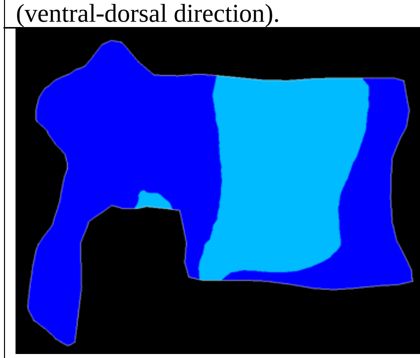
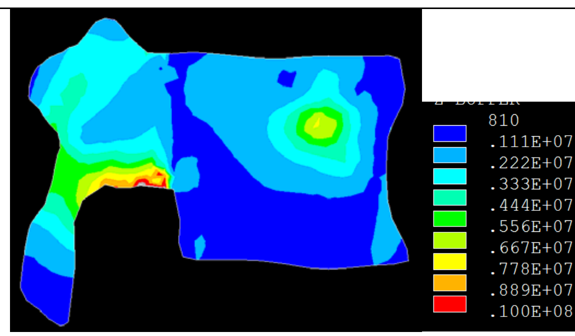
	
The Von Mises stress for the SIMload in the midline of vertebral body (medial-lateral direction).	The Von Mises stress for the MUSload in the midline of vertebral body (medial-lateral direction) (Grey areas above stress threshold).
	
The Von Mises stress for the SIMload in the midline of vertebral body (ventral-dorsal direction).	The Von Mises stress for the MUSload in the midline of vertebral body (ventral-dorsal direction).
	
The Von Mises stress for the SIMload cut through pedicle process and vertebral body (ventral-dorsal direction)	The Von Mises stress for the MUSload cut through pedicle process and vertebral body (ventral-dorsal direction)

Fig. (5). Von Mises stress for the SIMload and MUSload load cases.

30% larger von Mises stresses compared to the SIMload, but in certain areas in the vertebral body the stress distribution was even higher, namely in center and anterio-superior part, but also in the pedicles. These findings may be especially important for the analysis of elderly people since the risk of vertebral fractures in general increases with age [23] and fatigue lifetime of vertebral cancellous bone in particular decreases with age [24]. Furthermore, fatigue lifetime is highly dependent on the directional orientation between trabecular microstructure and acting loading [25]. Therefore even small changes in the orientation of the stress vector may result in a large decrease in fatigue lifetime and therefore increase the risk of fractures. The peak von Mises stresses were situated superio-anterior, which, together with the peak loading in the centre, would indicate development of wedge fractures as seen in the elderly population. Conclusively this indicates that it is essential to include muscle forces in spinal compression fracture risk prediction. However, at this point implementation of these results in clinical treatment and rehabilitation of osteoporotic fractures or fractures caused by a sport injury in the young healthy athlete are very complex and would have to be targeted specifically in future studies where the parameters described below have to be included.

In this study, the FE model did not include discs and the other vertebral bodies. Instead, the intervertebral discs were simulated as reactions forces. Earlier studies [26,27] by the authors have incorporated vertebral disks, but in the present case, the purpose of the study was merely to demonstrate the difference between exclusion and inclusion of the muscle forces.

In this study we included material data and geometric data of the 4th lumbar vertebrae and were able to simulate the resulting stress distribution for a lifting task using static analysis. The FE model was based on geometrical data from one test subject only and the stresses for any given individual could be significantly influenced by the general anatomical variation [28,29]. However, when analyzing the effects of geometric variation, as has been done in a previous study, by comparison of differences in contour points distances (CPD) in the segmentation process, it is shown, that it has a negligible role in the stress/strain distribution in the vertebral bodies [26]. Hence, fracture risk would be equally higher in the thoracolumbar region as for the lower lumbar region, when including muscle forces, since osteoporotic fractures often occur in the superior part of the vertebral bodies. Another important issue for fracture estimation is the bony material distribution. This could be implemented in a finite element model by using bone mineral content as material distribution data, which we have done in a previous study [26]. We measured 88 test subjects from L3 to L5 and found a large variation in between test subjects (70 MPa difference for 145 MPa young's modulus), but even as high a difference within the individual vertebral body, and it has been shown, that peak stresses increase by up to 74% in the osteoporotic vertebral body. Therefore for a comprehensive fracture analysis using the finite element method an adequate method of bony material estimation has to be included, and this will be undertaken in future studies.

Also, studies examining spinal loads indicate that much higher loads take place *in vivo* [30]. Furthermore, the full range of daily activities represented by reaction forces must be considered to be a simplified average estimate. However, to the best of our knowledge, though, there are no obtainable *in vivo* data on this subject.

CONCLUSION

The computations show quite clearly that muscle forces play a large and non-negligible role for the stress distribution in the vertebrae. Finite element models that neglect the muscle forces are likely to predict stress distributions that are quite far from reality. Ideally, FE models of the human vertebrae should include detailed muscle forces from a variety of activities of daily living and an e stimation of the material distribution of bone.

DISCLOSURE

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