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Determination of trunk muscle forces for flexion and extension by using a validated finite element model of the lumbar spine and measured in vivo data

Antonius Rohlmann*, Lars Bauer, Thomas Zander, Georg Bergmann, Hans-Joachim Wilke

Biomechanics Laboratory, Charité, Campus Benjamin Franklin, Hindenburgdamm 30, 12203 Berlin, Germany

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Abstract

Muscle forces stabilize the spine and have a great influence on spinal loads. But little is known about their magnitude. In a former in vitro experiment, a good agreement with intradiscal pressure and fixator loads measured in vivo could be achieved for standing and extension of the lumbar spine. However, for flexion the agreement between in vitro and in vivo measurements was insufficient. In order to improve the determination of trunk muscle forces, a three-dimensional nonlinear finite element model of the lumbar spine with an internal fixation device was created and the same loads were applied as in a previous in vitro experiment. An extensive adaptation process of the model was performed for flexion and extension angles up to 20° and −15°, respectively. With this validated computer model intra-abdominal pressure, preload in the fixators, and a combination of hip- and lumbar flexion angle were varied until a good agreement between analytical and in vivo results was reached for both, intradiscal pressure and bending moments in the fixators. Finally, the fixators were removed and the muscle forces for the intact lumbar spine calculated. A good agreement with the in vivo results could only be achieved at a combination of hip- and lumbar flexion. For the intact spine, forces of 170, 100 and 600 N are predicted in the m. erector spinae for standing, 5° extension and 30° flexion, respectively. The force in the m. rectus abdominus for these body positions is less than 25 N. For more than 10° extension the m. erector spinae is unloaded. The finite element method together with in vivo data allows the estimation of trunk muscle forces for different upper body positions in the sagittal plane. In our patients, flexion of the upper body was most likely a combination of hip- and lumbar spine bending.

Keywords: Lumbar spine; Muscle forces; Finite element method; Posture

1. Introduction

For biomechanical studies at the spine the loads are an essential factor. The spine is stabilized by muscle forces. Thus, the muscle forces are an important factor for the spinal loading (Crisco and Panjabi, 1991; McGill, 1992; Parnianpour et al., 1997; Rohlmann et al., 1998; Wilke et al., 1995). Muscle forces have been estimated for some standard exercises like standing, flexion and extension of the upper body, and lifting a load (Bergmark, 1989; Bogduk et al., 1992; Dolan and Adams, 1993; Guzik et al., 1996; MacIntosh et al., 1993; Nussbaum et al., 1995; Schultz and Andersson, 1981). Different approaches have been used in these studies. Some groups used mathematical models with and without EMG measurements. Calisse et al. (1999) and Zander et al. (2001) used the finite element method and results of in vivo load measurements on internal spinal fixators to predict muscle forces for standing and during flexion of the upper body. Both papers provide no results for extension of the lumbar spine.

*Corresponding author. Tel.: +49 30 8445 4732; fax: +49 30 8445 4729.
E-mail address: rohlmann@biomechanik.de (A. Rohlmann).

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Wilke et al. (2003) performed an extensive in vitro study with human cadaver lumbar spines. After applying the upper body weight in different flexion/extension positions, muscle forces were varied until the bending moment at the L1 vertebra was zero. Several loading combinations were studied and the intersegmental rotations, intradiscal pressures and loads on the internal fixators were measured. The results were compared with those of in vivo measurements. Mostly a good agreement could be achieved, except for flexion of the upper body where a significant difference in the fixator loads was ascertained. This means that for flexion of the upper body different loads acted in vivo and in vitro. In the in vitro experiment, intra-abdominal pressure was neglected. In addition, the instrumented fixators were fixed to the cadaver specimens in a nearly neutral position while the bridged region was distracted in the patients. Furthermore, lumbar flexion and hip flexion were studied separately in the in vitro experiment whereas in a patient probably a combination of both occurs.

The aim of this study was to determine the trunk muscle forces for standing, flexion and extension of the upper body. For this, the in vitro experiments performed by Wilke et al. (2003) were simulated in a validated finite element model of the lumbar spine and the uncertain loading parameters of the experiment (e.g. preload in the fixators, variation of muscle force direction) were adapted in the computer model until a good agreement between analytical and in vitro results is achieved. After this adaptation process, the analytical model predicted nearly the same global parameters as measured on average in the in vitro experiment. Then additional load modification was performed in order to achieve a good agreement between analytical and in vivo results. Finally, the trunk muscle forces were determined for the spine without a fixator for different flexion/extension angles. Using this way, the prediction of the muscle forces was improved and more reliable.

2. Methods

2.1. Previous experiment

In an in vitro experiment, Wilke et al. (2003) used seven fresh lumbar spine specimens from young donors. The mean age was 35 years (range 18–68). Before mounting them in a spine tester (Wilke et al., 1994) they were instrumented (i) with four pressure transducers for measuring intradiscal pressure, (ii) with an ultrasound-based motion analysis system for measuring intersegmental rotation, (iii) with several force transducers for measuring simulated muscle forces and spinal load and (iv) with a telemeterized internal spinal fixation device (Rohlmann et al., 1994, 1999, 2000b) for measuring the loads on the fixators. Lumbar flexion angles between -15° and 20° and hip flexion angles between 0° and 30° were studied. Ten different loading cases were studied by applying the upper body weight, a compressive follower load (Patwardhan et al., 1999; Rohlmann et al., 2001b), muscle forces and an abdomen supporting force (Calisse et al., 1999). For the measured parameters mean and range were determined and used for the comparison with the analytical results. The experiment is described in detail in their paper (Wilke et al., 2003).

2.2. Finite element model

A three-dimensional, nonlinear finite element model of the human osseoligamentous lumbar spine ranging from L1 to L5 was used (Zander et al., 2001) (Fig. 1). The element mesh of the vertebrae is based on the model used by Smit (1996). All seven ligaments were included in the computer model and represented by spring elements with nonlinear material properties. The material properties of all tissues were taken from the literature (Goel et al., 1995a, b; Shirazi-Adl et al., 1986) (Tables 1 and 2). The nucleus pulposus was simulated by an incompressible fluid-filled cavity. For the annulus fibrosus volume elements with superimposed spring elements were used. The latter represent the fibers and were mounted in four layers under alternating orientation of 30° and 150° to the mid-cross-sectional area of the disk. Fiber stiffness increased from the center to the outer shell. The facet joints had a gap of 0.5 mm and could transmit only compressive forces. The model was validated on the basis of experimental data from in vitro measurements (Rohlmann et al., 2001a, b), and a high degree of conformity was found. Pure moments in the three main anatomical planes were applied in these validation studies. A telemeterized bisegmental internal spinal fixation device (Rohlmann et al., 1994) spanning from L2 to L4 was implemented in the finite element model. The lower endplate of the L5 vertebra was rigidly fixed by constraining the corresponding nodes in the computer model. A compressive follower load (Patwardhan et al., 1999; Rohlmann et al., 2001b) was applied. This load was accomplished by forces of constant magnitude acting in the centers of adjacent vertebral bodies (Figs. 1 and 2). The finite element program ABAQUS and the pre- and postprocessor MSC/PATRAN were used. The highly nonlinear model allows large deformations and can simulate contact behavior at the facet joints. Furthermore, for ligaments and disks nonlinear material properties were chosen.

2.3. Loading cases

Of the 10 loading cases studied in the in vitro experiment the four most important ones were simulated

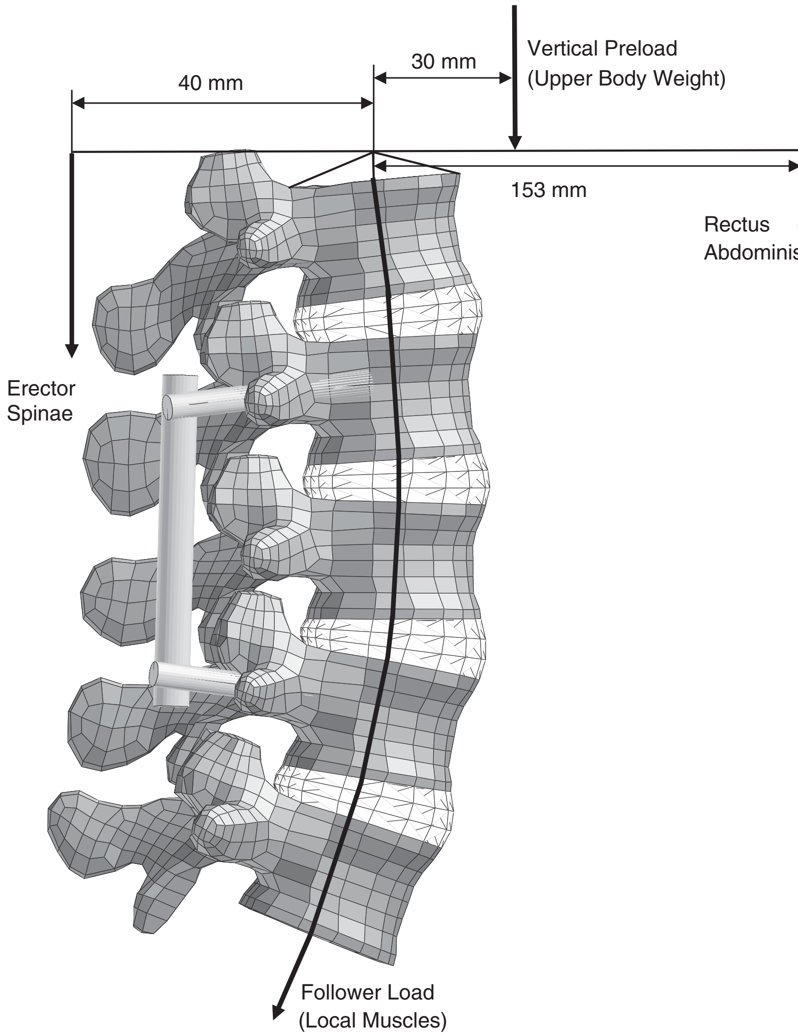

Fig. 1. Finite element model of the lumbar spine with the loads applied in the analysis.

Table 1
Material properties of different components

Material	Elastic modulus (MPa)	Poisson ratio (dimensionless)
Outer element layer of vertebral body (cortical/cancellous bone)	5000	0.3
Cancellous bone	500	0.2
Posterior bony elements	3500	0.25
Annulus fibrosus (ground substance)	3.15	0.45

in the finite element study. The weight of the upper body acts in the center of gravity, 200 mm cranial and 30 mm ventral of the T12/L1 disk center (Calisse et al., 1999; Stokes and Gardner-Morse, 1991). Flexion and extension occur in the sagittal plane. Thus, as in the experiment, only four muscle groups were simulated: left and right m. erector spinae, plus left and right m. rectus abdominus (Fig. 1). A compressive follower load (Patwardhan et al., 1999; Rohlmann et al., 2001b) was applied to substitute for the unknown stabilizing effect of local muscles.

Two kinds of flexion angles were studied:

(a) Lumbar flexion angles between -15° (extension) and $+20^\circ$ (flexion) in 5° steps.
(b) Hip flexion angles between 0° and 30° varied in 10° steps without bending the lumbar spine. Thus, no flexion or extension was superimposed. In the finite element model, hip flexion was simulated by rotating the whole model around the hip center.

The following four load cases (Fig. 1) were investigated to achieve the different flexion angles:

(1) Vertical force of 220 N (representing upper body weight) (load case 2 in the in vitro study of Wilke et al., 2003).
(2) Vertical force of 260 N plus follower load of 200 N (load case 5 in the in vitro study).

Table 2
Ligament stiffnesses for three different strain ranges

Ligaments	Stiffness (N/mm)	Strains between (%)	Stiffness (N/mm)	Strains between (%)	Stiffness (N/mm)	Strains higher than (%)
Lig. long. anterius	347	0–12.2	787	12.2–20.3	1864	20.3
Lig. interspinalia	1.4	0–13.9	1.5	13.9–20	14.7	20
Lig. capsulae	36	0–25	159	25–30	384	30
Lig. intertransversia	0.3	0–18.2	1.8	18.2–23.3	10.7	23.3
Lig. flava	7.7	0–5.9	9.6	5.9–49	58.2	49
Lig. long. posterius	29.5	0–11.1	61.7	11.1–23	236	23
Lig. supraspinalia	2.5	0–20	5.3	20–25	34	25

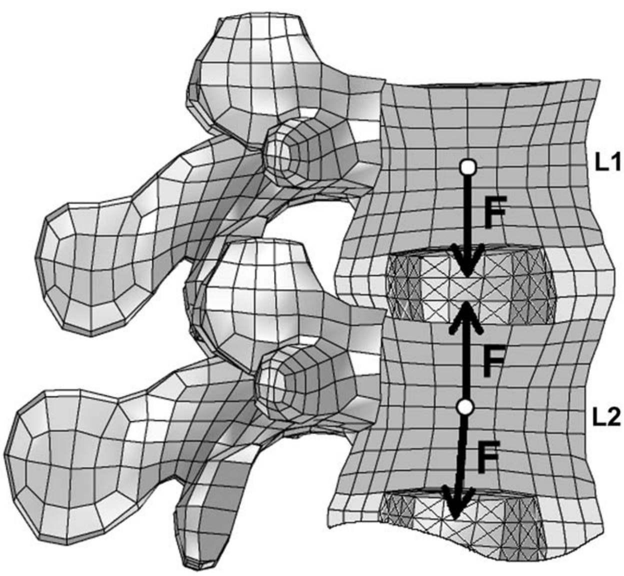

Fig. 2. Representation of the follower load F for the vertebrae L1 and L2. Only the left half of the vertebral bodies and disks is shown. The forces in the center of a vertebral body always point toward the adjacent vertebral body centers and are normally not in line since the follower load acts tangential to the spinal curvature.

(3) Vertical force of 260 N plus follower load of 200 N plus total force in the m. rectus abdominus of 50 N (simulating muscle co-contraction) (load case 6 in the in vitro study).
(4) Vertical force of 260 N plus follower load of 200 N plus total force in the m. rectus abdominus of 100 N (simulating muscle co-contraction) (load case 7 in the in vitro study).

Thus, the results for analytical and experimental results are compared for 44 different loading cases.

As in the previous experiment (Wilke et al., 2003), a pure moment was first applied on the superior endplate of the L1 vertebra to bring L1 in a predetermined rotational position (e.g. 20° lumbar flexion). Then the forces of one of the four loading cases were applied without changing the rotation of the L1 vertebra. Finally, the force representing the m. erector spinae was varied until no additional bending moment on the superior endplate of the L1 vertebra was necessary to keep the spine in the desired rotational position (e.g. 20° lumbar flexion).

2.4. Evaluation of the results

For comparison with the experimental results intersegmental rotations, intradiscal pressures, bending moments in the fixators and total muscle forces in the erector spinae were calculated. In vivo intradiscal-pressure results are available for standing, 20° flexion and 15° extension (Wilke et al., 1999). Fixator loads were compared with those measured in three patients operated due to degenerative instability (Rohlmann et al., 1999, 2000a, b). Anterior interbody fusion using bone grafts was performed in these patients about 3 weeks after implantation of the fixators. Only loads measured prior to anterior interbody fusion were chosen for comparison with the data of the finite element analysis. In vivo fixators loads are available for standing, 15° flexion and 15° extension. No biomechanical in vivo data could be found in the literature for hip flexion only and for intersegmental rotation in the lumbar spine after insertion of internal fixators.

2.5. Modifications in the computer model

Four steps were performed to estimate trunk muscle forces.

(1) Analysis of the initial situation
The fixators were fixed to the model without any preload and finite element analyses were performed for the 44 different loading conditions.
(2) Adaptation of the computer model to the in vitro situation
In such a complex experiment as that of Wilke et al. (2003), it is not possible to control all parameters very precisely (e.g. changing direction of muscle cables during loading) since all measurements

on a specimen had to be performed on 1 day. Thus, the uncertain parameters were varied slightly in the computer model. The distance of the internal fixators to the lumbar spine as well as the preload on the fixators due to tightening the screws varied from specimen to specimen. In the finite element model, the distance of the pedicle screws to which a fixator was clamped, was varied by 1 mm thus causing different axial preloads on the bridged region. In the experiment, half of the L1 vertebra was embedded and a plate was fixed to the top of the vertebra. The plate represented the cross-sectional area of the human body at that level and served for application of the body weight and muscle forces. Small deviations of the center of the L1 vertebra relative to the plate were unavoidable. Therefore, the position of the L1 vertebra relative to the plate was modified in a–p direction in the computer model. Since the load direction of the muscles varied slightly for the different specimens in the experiment, the direction of the muscle forces were modified up to 5°.

(3) Adaptation of the computer model to the in vivo situation

In the in vivo experiments, an intra-abdominal pressure was probably acting (Marras and Mirka, 1996). The patients for which the loads of the fixators were chosen suffered from degenerative disk disease. Therefore, the bridged region was distracted during surgery until normal disk height was reached. In order to get the same bending moments in the fixators as in the in vivo experiment, the longitudinal rods of the fixators in the finite element model had to be slightly elongated. It was further assumed that the in vivo motion of a patient consists of a combination of both hip- and lumbar flexion. Therefore, the influences of the intra-abdominal pressure, preload in the internal fixators and combinations of lumbar- and hip flexion were studied to adapt the model to the in vivo situation. Intra-abdominal pressure was simulated by applying a distributed force of 30 N per vertebra in the mid-sagittal plane on the ventral side of the spine. The preload in the fixators was adopted by increasing the distance of the pedicle screws in a standing position. For the flexion/extension angles where in vivo data are available different combinations of hip- and lumbar flexion were studied.

(4) Muscle forces for the healthy lumbar spine without fixators

The muscle forces for a healthy spine are more important than for a spine with a fixator. Thus, in a final step, the fixators were removed from the finite element model and the trunk muscle forces determined in the same way as described above. The muscle forces in the erector spinae were estimated at angles between 20° extension and 30° flexion in steps of 5° for the two loading cases which results are closest to those of the in vivo study.

3. Results

3.1. Initial computer model

A reasonable agreement of the analytical and in vitro results could be achieved for the intradiscal pressure. For each loading case the difference was less than 0.2 MPa that is 14% of the maximum value. Reasonable agreement was also found for intersegmental rotation during lumbar flexion/extension. The difference was mostly less than 2°. However, for hip flexion the results differed strongly (up to 5.5°). Great differences were also found for resulting bending moments in the internal fixators (up to 2.5 Nm that is 48% of maximum value) and for the forces in the m. erector spinae (up to 320 N, that is 43% of maximum value).

3.2. Comparison of analytical and in vitro results

After slightly modifying preload in the fixators and direction of the muscle forces a good agreement could be achieved for intersegmental rotation at L4/L5 (Fig. 3) and intradiscal pressure in disk L4/5 (Fig. 4). Good agreement was also achieved at the other segments. For all 44 loading cases, the calculated values for intradiscal pressure and intersegmental rotation were within the range measured in the seven specimens in the in vitro experiment. The average difference between analytical and in vitro results for intersegmental rotation was 1.18° (SD: 0.89°) and for intradiscal pressure 0.09 MPa (SD: 0.05 MPa). Related to the maximum value the deviations were 10.3% and 6.6%, respectively. For comparison, in seven specimens the intersegmental rotation measured for a hip flexion angle of 30° varied between −2.9° and +3.8° which is 58.3% related to the absolute maximum value measured. The resultant bending moments in the fixators (Fig. 5) also showed a reasonable agreement between the finite element analysis and the in vitro study. Mostly the calculated values were within the range of the measured bending moments. The average difference was 0.82 Nm (SD: 0.49 Nm) which is 16.2% related to the maximum value. The total forces in the m. erector spinae agreed well for the two different methods. The average difference was 52.9 N (SD: 31.5 N; 6.0% of max). Variation of the other parameters had only a slight effect.

3.3. Comparison of analytical and in vivo results

Using the validated finite element model, simulation of the intra-abdominal pressure had only a minor

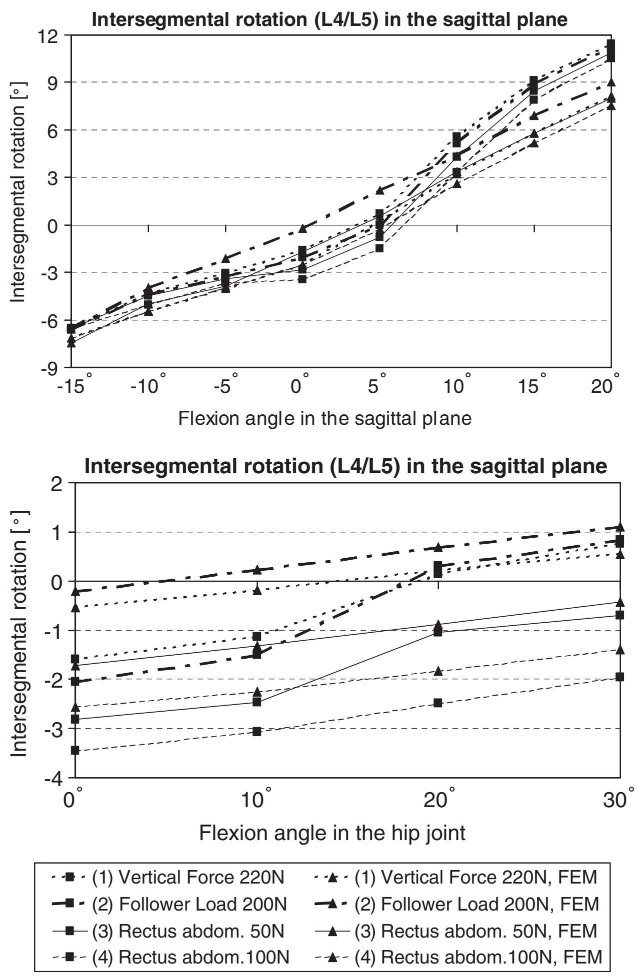

Fig. 3. Comparison of calculated and in vitro measured intersegmental rotation at L4/L5 for lumbar flexion/extension. Intersegmental rotations are shown for flexion of the lumbar spine (top) and flexion in the hip joint (bottom). Quadratic symbols represent experimental results, triangular symbols analytical values.

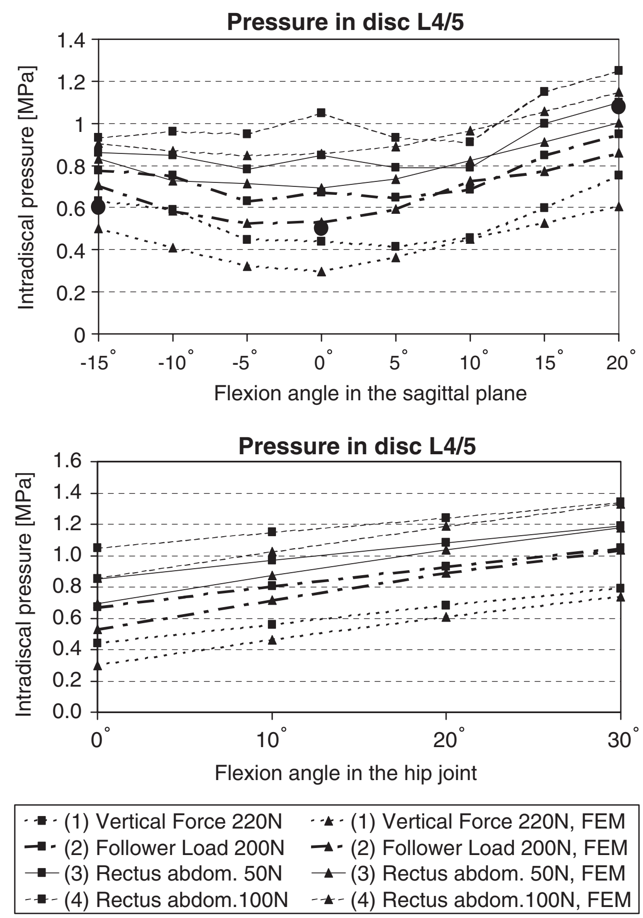

Fig. 4. Comparison of calculated and in vitro measured intradiscal pressure in disk L4/5 for lumbar flexion/extension. Disk pressures are shown for flexion of the lumbar spine (top) and flexion in the hip joint (bottom). Quadratic symbols represent experimental results, triangular symbols analytical values and the three round symbols (top) in vivo results.

influence on both parameters measured in vivo, intradiscal pressure and bending moments in the fixators (less than 0.1 MPa and less than 0.1 Nm, respectively). After distracting the region bridged by the fixators, as performed in the patients, the loading case 2 led to a good agreement of calculated and in vivo measured intradiscal pressure (Fig. 6) and bending moments in the fixators (Fig. 7) for standing. For flexion, good agreement was achieved only when the flexion occurred to 2/3 in the hip joint and 1/3 in the lumbar spine. The agreement was best for the loading case 2 and second for the loading case 3 (Figs. 6 and 7). The curves suggest a force of about 20 N in the m. rectus abdominus for 20° flexion. For 15° extension a motion proportioning of 1/3 in the hip joint and 2/3 in the lumbar spine led to a good agreement of analytical and in vivo results (Figs. 6 and 7). Also for extension best agreement was achieved for loading case 2. An additional force of 15 N in the m. rectus abdominus leads to perfect agreement of the results for 15° extension. For standing, flexion and extension, the maximum differences between loading case 2 and in vivo results were 0.08 MPa and 0.23 Nm, respectively.

3.4. *Trunk muscle forces for the healthy spine without fixators*

For different flexion/extension angles in the sagittal plane the forces in the m. erector spinae are shown in Fig. 8. The results represent the situation of an intact lumbar spine without an internal fixator. The values for the loading cases 2 and 3 are presented. For flexion of the upper body there is nearly a linear relationship between the force in the m. erector spinae and the flexion angle. A force of 50 N in the m. rectus abdominus increases the force in the m. erector spinae between 100 and 150 N. According to Figs. 6 and 7 a force up to 20 N acts in the m. rectus abdominus. For standing a muscle force of 170 N is predicted and for 30°

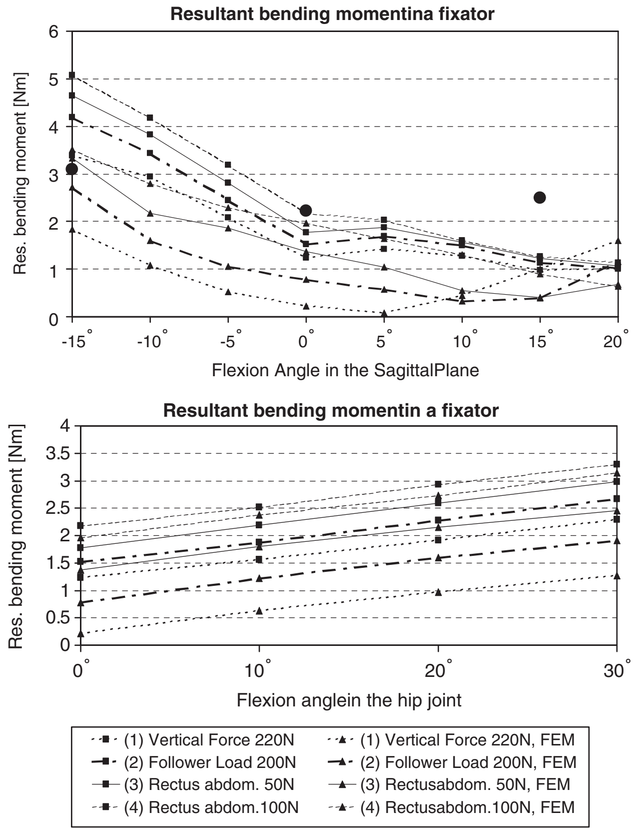

Fig. 5. Comparison of calculated and in vitro measured resultant bending moments in the fixators for lumbar flexion/extension. Bending moments are shown for flexion of the lumbar spine (top) and flexion in the hip joint (bottom). Quadratic symbols represent experimental results, triangular symbols analytical values and the three round symbols (top) in vivo results.

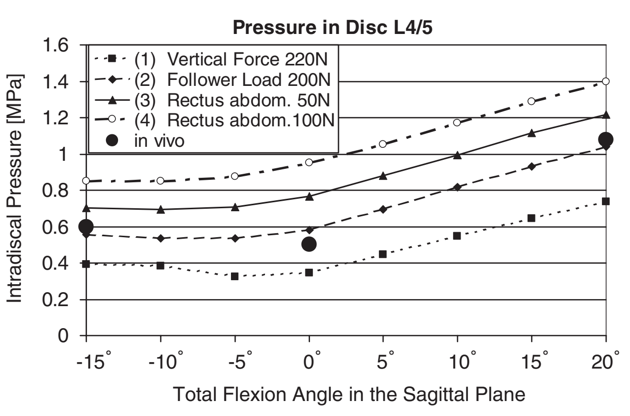

Fig. 6. Comparison of calculated and in vivo measured intradiscal pressure for various flexion angles in the sagittal plane.

flexion a force of about 600 N when a force of 20 N is assumed in the m. rectus abdominus. The m. erector spinae is unloaded for extension angles greater 10°.

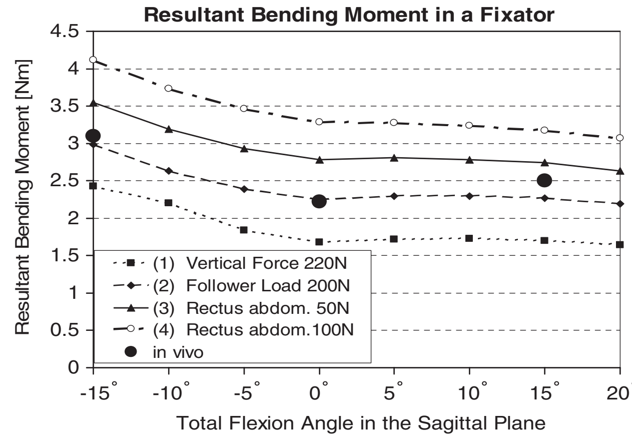

Fig. 7. Comparison of calculated and in vivo measured resultant bending moments in the fixators for different flexion angles.

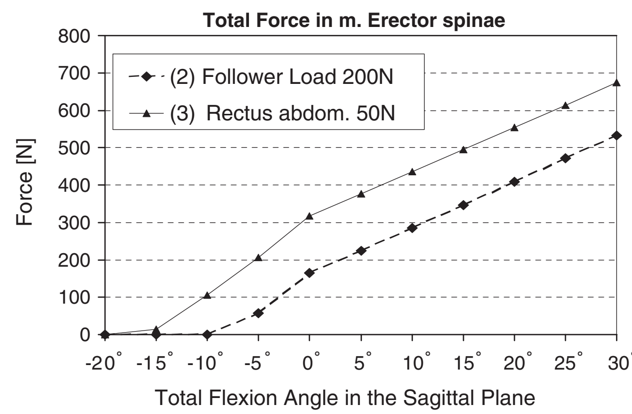

Fig. 8. Forces in the m. erector spinae for different body positions in the sagittal plane calculated for the lumbar spine without an internal fixator.

4. Discussion

In order to determine trunk muscle forces for motion in the sagittal plane, a finite element model of the lumbar spine was used. By adapting the preload in the internal fixation devices and the direction of the trunk muscle forces a good agreement between calculated intersegmental rotations, intradiscal pressures and loads on the fixators and those values measured in vitro could be achieved for 44 different loading cases. A realistic simulation of the in vivo situation in a computer model became feasible by distracting the bridged region as was done in the patients and by combining hip- and lumbar flexion. With this validated model, the loading situations for flexion, standing and extension were realistically determined in a finite element study for the first time.

As in all finite element studies involving the spine, several simplification and assumption had to be made. Although a good agreement between experimental and analytical results was achieved, e.g. the stresses in the different structures and the contact forces in the facet joints may be not very precise. But this has no disadvantages for the results in this study.

Great interindividual differences in the spinal loads may occur. Body weight, body size and posture vary strongly and will have an influence on the muscle forces. Thus, our calculated results represent values lying within a possible range. Since median values were used in the in vitro study, the calculated results should describe a representative person.

Calisse et al. (1999) using a geometrically strongly simplified finite element model of the lumbar spine (the vertebral bodies were modeled as cuboids) predicted a force in m. erector spinae of 250 N for standing and 1000 N for 60° flexion. The value for standing is 80 N higher than that of the present study. They did not determine the forces for extension or for other flexion angles than 60°.

Zander et al. (2001) estimated trunk muscle forces for standing and up to 30° flexion. They assumed local muscle forces which represented m. multifidus, m. iliocostalis lumborum pars lumborum, m. longissimus thoracis pars lumborum and m. longissimus thoracis pars thoracis. Depending on the amount of the local muscle force they predicted a force in the m. erector spinae up to 300 N for standing and between 500 and 800 N for 30° flexion. Our values lay within these ranges. Zander et al. (2001) did not determine muscle forces for extension of the upper body.

In their experimental study, Wilke et al. (2003) predicted forces in the m. erector spine of 100 N for standing, 130 N for 15° extension, and 520 N for 30° flexion of the upper body. The values for standing and 30° flexion are slightly lower than those determined in the present study but their value for 15° extension is higher. The experimental results, however, represent a lumbar spine with a bisegmental internal fixation device.

Applying a follower load instead of a great number of small forces simulating the local dorsal muscles makes realistic loading in in vitro studies feasible. It seems that for sagittal plane motion few global muscle forces are sufficient to achieve realistic results. Our results as well as the results of a previous in vitro study (Wilke et al., 2003) showed that the follower load is a suitable tool to adjust the intradiscal pressure to physiological values without significantly affecting intersegmental motion. However, the local muscles act at a larger lever arm to the center of rotation than a follower load and thus have a greater stabilizing effect. Therefore, local muscles contribute somewhat more to the equilibrium of spinal moments than the follower load. Thus, replacing local muscles by a follower load leads to slightly too high intradiscal pressures. The body weight and the follower load have the greatest influence on intradiscal pressure. In our study, we assumed a vertical preload of 260 N which corresponds to a body weight of about 56 kg. A higher body weight would require a lower follower load to achieve the desired intradiscal pressure but would increase the force in m. erector spinae necessary to zero the moment at the cranial endplate of L1 during flexion. In reality, the weight of the upper body is partly carried by the pelvis.

Intra-abdominal pressure does not have a very strong influence on spinal load and muscle forces for the loading cases studied. Thus, it can normally be neglected. However, when lifting heavy loads, intra-abdominal pressure may affect muscle forces.

In a previous in vitro experiment (Wilke et al., 2003), hip- and lumbar flexion were studied separately. Additional measurements would have been possible only with specimens of less quality due to long testing time. The finite element method allows the study of unlimited loading cases without altered material properties.

A low antagonistic force in the m. rectus abdominus was predicted for the loading cases studied. Increasing this force considerably increases the force in the m. erector spinae and thus the spinal load.

The m. erector spinae is necessary to initiate an extensional movement, but for the steady state, the upper body weight may have its point of application posteriorly to the lumbar spine, thus resulting in an extension bending moment. This moment reduces the necessary force in the m. erector spinae, sometimes even to zero.

In our model, only one global trunk muscle was assumed anterior and posterior of the spine. In reality, a larger number of muscles are involved in flexion and extension of the upper body. Thus, each predicted muscle force represents the action of several muscles and the real forces in the erector spinae and the rectus abdominus are lower. In experimental studies, the application of this few muscle forces is feasible (Wilke et al., 2003).

Trunk muscle forces can only be determined indirectly because a direct measurement is not possible. In this finite element study, data from an extensive in vitro experiment were used to validate a computer model and in vivo data measured in patients were employed to estimate trunk muscle forces for flexion and extension of the upper body. Only the combination of different methods led to reliable results since the strengths of the different methods could be merged.

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