# NUMERICAL SIMULATION AND PARAMETER IDENTIFICATION OF HUMAN LUMBAR FSUs IN HYDROTRACTION

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**Abstract: Numerical analysis and parameter-identification of human lumbar spine segments are presented for centric tension, to simulate the behaviour of human lumbar spine segments during traction hydrotherapy. The finite element models of the functional spinal units (FSUs) are created from the results of a large-scale in vivo experimental parameter-analysis during traction hydrotherapy, a special Hungarian invention. Global elongations of lumbar spine segments were measured as the change of the distance between two adjacent spinous processus, by using a special subaqual ultrasound measuring method. Tensile elastic moduli and viscoelastic parameters of segments have been derived from the measured time-related deformations, in terms of sexes, age and other biomechanical parameters. 2D and 3D FEM models were created, and the global elongations of FSUs were used for identification of the local material parameters of the component organs of lumbar spine segments.**

## Introduction

Spinal deformations have been measured during the usual traction bath therapy by *Kurutz et al.* [1,2] by developing a special subaqual ultrasound method for measuring the deformations of lumbar segments of human body suspended in water when the effect of muscles can practically be excluded. Elongation of segments was considered as the change of the distance between the spinous processes of adjacent vertebrae. Time-related in vivo deformations of segments L3-L4, L4-L5 and L5-S1 were documented. More than 3000 ultrasound pictures of 409 segments of 155 adult patients have been evaluated. Based on the measured time related spinal deformations in terms of aging, sexes, body weight and height, parameter-dependent global elastic and viscoelastic numerical models of human lumbar-lumbosacral spine segments of part L3-S1 were created for tension [3,4]. The numerical models were used for numerical simulation of traction therapies and for parameter identification of the component organs of lumbar segments.

Finite element softwares have been used for numerical simulation of the lumbar spine in traction therapy. The numerical model of the spine segment is based on the anatomic model of the functional spinal unit (FSU). The anatomic model of the FSU concern a single lumbar motion segment consisting of two vertebrae with the disc between them, and the surrounding ligaments.

## Materials and Methods

The suspension hydrotherapy is a Hungarian invention. The therapy needs a specially deepened basin with elastic suspension equipment. In our experiments, cervical suspension has been applied exclusively, since this mode of support provides the most effective stretching load to the lumbar part of spine, moreover, in this case the effect of muscles can be practically neglected. The extra lead loads have been applied on ankles.

In the case of cervical suspension, three load effects cause traction deformations along the spinal column: (1) the decompressive force as the removal of the compressive load of the body weight existing before the treatment, (2) the active tensile force due to the buoyancy, and (3) the extra loads applied for the therapy. By comparing the three component forces detailed in [4], if applying cervical suspension, the dominant stretching load is the decompressive force. It takes about 97% of the stretching load if no extra loads are applied. In the case of free cervical suspension in water, the effect of muscles can be neglected. Indeed, the free elastic suspension of the body and the gentle lukewarm water together with the feeling of unloading of the body help to relax muscles of the nearly sleeping patients.

Tensile load effects depend on the definition of the tensile deformations. By definition, tensile deformations of segments suspended in water are considered as the decrease of compressive deformations existing before the treatment. Tensile deformation of segments has been considered as change of the distance between two spinous processes of neighboring vertebrae.

Elongation values of lumbar segments L3-4, L4-5 and L5-S1 have been measured. More than 400 lumbar segments of 155 patients have been measured, more than 3000 ultrasound pictures have been evaluated. The 155 measured subjects have been divided into two groups: 88 patients of less degenerated segments with medically prescribed extra lead loads, that is, 20-20 N applied symmetrically on the ankles, and a group of 67 patients with more degenerated segments without any extra loads. This latter group consisted of patients to whom the extra lead loads have been contraindicated due to certain degeneration of the lumbar segments or


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discs. Generally, deformations have been measured in four time-phases of the treatment: just before the treatment; immediately at being suspended in the water; some times after; at the end of the 20 minutes long suspension.

Tensile deformability of lumbar segments was measured and evaluated in terms of biomechanical parameters, like sex, aging, body weight and height and body mass index.

The numerical model of the spine segment is based on the anatomic model of the functional spinal unit (FSU). The anatomic model of the FSU concern a single lumbar motion segment consisting of two vertebrae with the disc between them, and the surrounding ligaments.

The vertebrae consist of an anterior block of bone, named vertebral body, and a posterior bony ring, known as the neural arch, containing articular, transverse and spinous processes. The vertebral body is a roughly cylindrical mass of cancellous or spongy bone contained in a thin shell of cortical bone. Its superior and inferior surfaces are the slightly concave vertebral end-plates.

The intervertebral disc is composed of three distinct parts: the nucleus pulposus, the annulus fibrosus and the cartilaginous end-plates. The nucleus pulposus has a centrally located area composed of a very loose jelly-like material. It is highest in young age and tends to decrease with aging. The annulus fibrosus surrounds the nucleus like a ring. This structure is composed of fibrous tissue in concentric laminated bands. The skew fibers have opposite direction in any band. The cartilaginous end-plates separate the disc from the vertebral body, that is replaced by bone with aging.

The ligaments are uniaxial structures, they can carry loads along the direction of them. They resist tensile forces but buckle when subjected to compression. In this aspect, they are much like rubber bands.

Since in the suspension traction hydrotherapy, the effect of muscles can be excluded, in our model, no muscle effects are considered.

In the numerical simulation three type of finite element models were used: a simple 2D model, a refined 2D model and a refined 3D model.

As the first step of the numerical simulation, we wanted to check the measured elongations of FSUs. For this program, the simple 2D model was used. Since the healthy spine segment is a symmetrical structure, the simple 2D model was related to the sagittal plane of the FSU. For the material constants we used the results of *Antosik et al.* [5] and *Ciach et al*. [6,7]. The geometrical data of the FSU were obtained by the measures of a typical lumbar segment. The support of the structure was along the inferior surface of the lower vertebra. The load was distributed along the superior surface of the upper vertebra.

For the global behaviour of the segment complex, we supposed bilinear material law. We assumed that the decompression takes place by the large stiffness $E_1$ that is valid for compression, and for the active tension of buoyancy and for the extra weights, we supposed a smaller stiffness $E_2$=$E_3$, respectively. This behaviour has been simulated by applying different disc moduli both for compression and tension.

As the second step of the numerical simulation, we refined the 2D FEM model to simulate the viscoelastic creep process, by applying NASTRAN software. For viscoelastic numerical model, the Poynting-Thomson type three-parameter spring-dashpot models were used, where spring constants represent the elastic properties, damping coefficients concern creep effects. The three parameters of the numerical creep model, namely, the spring constants and the damping coefficient were determined by the time-related in vivo measurements. Deformations have been measured in three time instant, from which the parameters could be calculated, by supposing that the last state at the $20^{th}$ minute of the treatment concerns the steady state of the creep process. Creep functions of the lumbar segments have been obtained in terms of sexes, aging and other parameters. From the results of the measured lumbar segments L3-4, L4-5 and L5-S2, a general lumbar FSU model have been created for numerical simulations.

For parameter identification, the NASTRAN model has been improved to the 3D configuration of the lumbar FSU. The segment model has 35 000 nodal points with the relating 110 000 equations.

In parameter identification, the in vivo measured global elongations of lumbar segment complex are used as the control parameters in determining the local behaviour of the component organs. Thus, in identification of the material parameters of the component organs, the global elongations are used in the loading process by using the created finite element models. The parameter identification aims to determine the in vivo controlled tensile behaviour and tensile material moduli of the component organs of lumbar segments that are completely missing in the international literature.

During the parameter identification process, the material moduli of certain organs are kept constant while the material parameters of other organs are considered to be the variable of the problem. Under these parameters, numerical simulation is investigated, and those results are considered to be realistic and are accepted that leads to the in vivo measured elongations. The intervals of material moduli used in parameter identification are seen Table 1.

Table 1: Intervals of material parameters in identification

| **Organs** | **Young's Modulus MPa** |
|---|---|
| cortical bone | 1000-20 000 |
| trabecular bone | 50-1000 |
| disc annulus fibrosus | 5-500 |
| disc nucleus pulposus | 1 |
| ligaments | 10-100 |


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## Results and discussion

Weightbath therapy seems to be evident to analyze lumbar traction effects in pure centric tension without the influence of muscles. *Andersson et al* [8] concluded that by active traction in laying posture, the back muscles contract and the disc pressure increases. Similarly, *Ramos and Martin* [9] verified the inverse relationship between the applied traction load and the intradiscal pressure. These observations verify the importance of the traction bath hydrotherapy. Namely, by applying cervical suspension in water, the contracting effect of muscles can be neglected.

At the end of the 20 minutes long treatment, the total mean tensile deformation of the lumbar FSU model L3-S1 without extra weight is about 0,7-0,9 mm, while with extra weights it is about 0,8-1,4 mm. These results correspond to the results of *Chen et al* [10] for traction therapy, by obtaining that the intervertebral disc distance increased by 1,34 mm for prolapsed, and by 0,87 mm for normal discs.

It has been numerically verified that deformations decrease with increasing age of patients in both the instant elastic and the time-dependent creeping phases of the traction therapy. These observations support the results of *Acaroglu et al* [11] who studied human lumbar annulus fibrosus to evaluate the effects of aging and degeneration on the tensile properties. We have observed that not only the ratio of rigidity, but also the effect of aging is more significant in distal direction. Similarly, *Adams et al* [12] showed that the effects of age and degeneration were greater at L4-L5 than at L2-L3.

Viscoelastic tensile deformability in relation of sex and aging is seen in Table 2. and 3. for the created general lumbar FSU model L3-S1 of less degenerated segments, by distinguishing sex and aging, respectively. Three age classes are considered: a young class under 40, a middle-aged class between 40-60, and an old class over 60 years. Since the traction load was constant during the 20 minutes long treatment, a typical creeping process developed and could be documented numerically.

Table 2: Sex-dependent parameters of the creep model of general lumbar FSU model L3-S1

| extra weight 20-20 N | **units** | **male** | **fem** | **total** |
|---|---|---|---|---|
| segments | *numb* | 108 | 128 | 236 |
| mean age | *years* | 47,9 | 51,7 | 50,0 |
| mean weight | *N* | 758 | 671 | 711 |
| mean height | *cm* | 175 | 164 | 169 |
| mean BMI | *kg/m²* | 24,9 | 25,1 | 25,0 |
| elong. at t=0 min | *mm* | 0,66 | 0,43 | 0,52 |
| elong. at t=3 min | *mm* | 0,83 | 0,74 | 0,78 |
| elong.at t=20min | *mm* | 1,15 | 1,11 | 1,13 |
| Creep parameters | | | | |
| spring coeff. $c_1$ | *N/mm* | 742 | 1018 | 887 |
| spring coeff. $c_2$ | *N/mm* | 999 | 643 | 756 |
| damping coeff. k | *Ns/mm* | 422 | 190 | 245 |
| time constant T | *min* | 7,04 | 4,93 | 5,40 |

Table 2 illustrates the creep moduli of the general lumbar FSU model for numerical purposes, by distinguishing the sexes. The mean initial elongation of women is about the 66% of the deformation of men. However, the final deformations are quasi equal, since the damping of men is more than double that of the women. The first spring coefficient is equal to the initial tensile stiffness. Initial tensile stiffness of males is about 750 N/mm, for females it is about 1000 N/mm. The damping coefficient of males is about 400 Ns/μm, for females it is about 200 Ns/μm. Consequently, after an initially smaller deformability of females, larger creep elongations occur. The final viscoelastic elongation is quasi equal.

Creep behaviour and creep curves of the general lumbar segment model L3-S1 are significantly different in terms of aging, as seen in Table 3. Taking three age-classes into consideration, it was numerically verified that the initial stiffness is increasing, consequently, deformability is decreasing with aging,. Similarly, damping coefficients also increase, thus, creep deformability decrease with aging. This tendencies are valid for both sexes.

Table 3: Age-dependent parameters of the creep model of general lumbar FSU model L3-S1

| extra weight 20-20 N | **units** | **under 40** | **40-60** | **over 60** |
|---|---|---|---|---|
| segments | *numb* | 35 | 161 | 40 |
| mean age | *years* | 26,5 | 50,7 | 67,5 |
| mean weight | *N* | 713 | 721 | 670 |
| mean height | *cm* | 176,7 | 169,0 | 160,3 |
| mean BMI | *kg/m²* | 22,9 | 25,2 | 26,0 |
| elong. at t=0 min | *mm* | 0,94 | 0,52 | 0,25 |
| elong. at t=3 min | *mm* | 1,25 | 0,78 | 0,39 |
| elong.at t=20min | *N/mm* | 1,51 | 1,19 | 0,55 |
| tens.stiff. t=0 min | *N/mm* | 492 | 899 | 1748 |
| tens.stiff. t=20 min | *mm* | 306 | 393 | 794 |
| Creep parameters | | | | |
| spring coeff. $c_1$ | *N/mm* | 492 | 899 | 1748 |
| spring coeff. $c_2$ | *N/mm* | 812 | 698 | 1456 |
| damping coeff. | *Ns/mm* | 186 | 256 | 417 |
| time constant T | *min* | 3,82 | 6,11 | 4,77 |

In Fig. 1 the creep curves of the general lumbar segment model L3-S1 is illustrated in terms of aging and sex.

We have found both the initial stiffness and the damping moduli increase, consequently, both the elastic and viscoelastic deformability decrease in distal direction, Similar effect can be observed in terms of aging, seen in Fig. 1a, namely, both the initial stiffness and the damping moduli increase, consequently, both the elastic and viscoelastic deformability decrease with aging. As for the effect of sex, seen in Fig. 1b, after an initially smaller deformability of females, due to the smaller damping of women, larger creep elongations occur. The final viscoelastic elongations are equal


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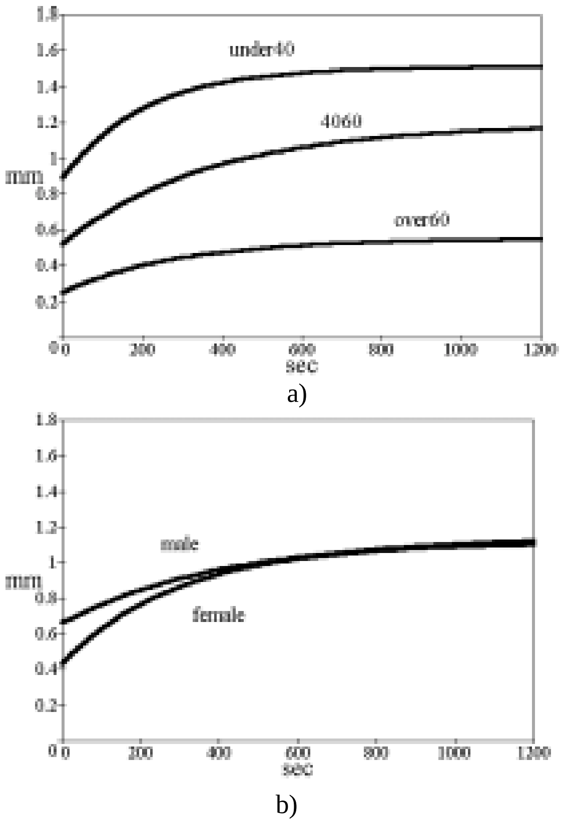


Figure 1: Creep functions of the general lumbar FSU model L3-S1 in terms of a) aging and b) sex

As a result of the numerical simulation, Fig 2. illustrates the effect of Young's moduli of disc annulus on the global elongation of FSU, by keeping constant the Young's moduli of trabecular bone as $E_2$=100 MPa, of ligaments as $E_4$=10 MPa and of disc nucleus as $E_5$=1 MPa. The results show that the Young's modulus of the annulus has a significant effect to the deformability of segments, and we can suppose that the tensile modulus of elasticity of the annulus is far smaller than the compressive one.

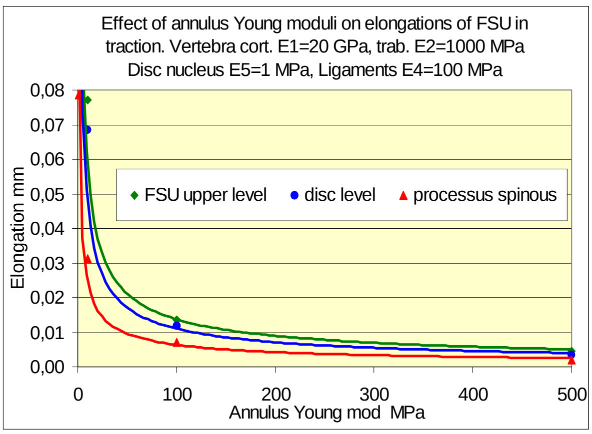


Figure 2: The effect of Young's moduli of disc annulus on the elongation of the segment

This result seems to be evident since the basic property of the annulus is to resist to compression. Another observation that changing the stiffness of the cortical bone has no significant effect on the deformability of segments.

In Fig 3. the effect of Young's moduli of disc annulus is illustrated on the vertical stresses of the component organs: cortical and cancellous bone, internal and external annulus, by keeping constant the material moduli as follows: Young's moduli of trabecular bone $E_2$=100 MPa, of ligaments $E_4$=10 MPa and of disc nucleus $E_5$=1 MPa. The results show that the Young's modulus of the annulus has practically no effect on the occurring stresses in the cortical or trabecular bone, however, it has a significant effect on the stresses of the nucleus and annulus itself.

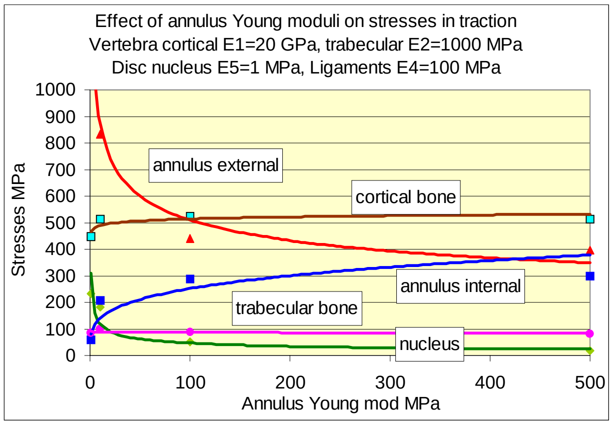


Figure 3: The effect of Young's moduli of disc annulus on the stressesof the component organs of segment

In stress analysis the mean vertical stresses in certain horizontal sections of FSU were examined for each organ in terms of the Young's mudulus of the component organs. In Fig. 4. and 5. the coupled effect of the soft tissues, disc annulus and ligaments are illustrated on the stresses of disc annulus and nucleus, by keeping the Young moduli of the cortical and cancellous bone of vertebrae constant as E1=20 Gpa and E2=1000 Mpa, respectively.

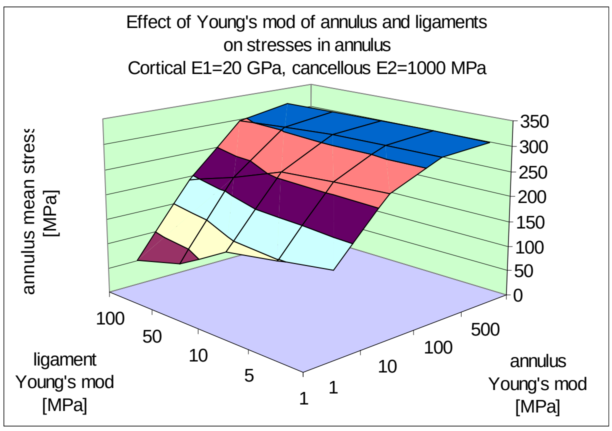


Figure 4: The coupled effect of Young's moduli of disc annulus and ligaments on the stresses of disc annulus


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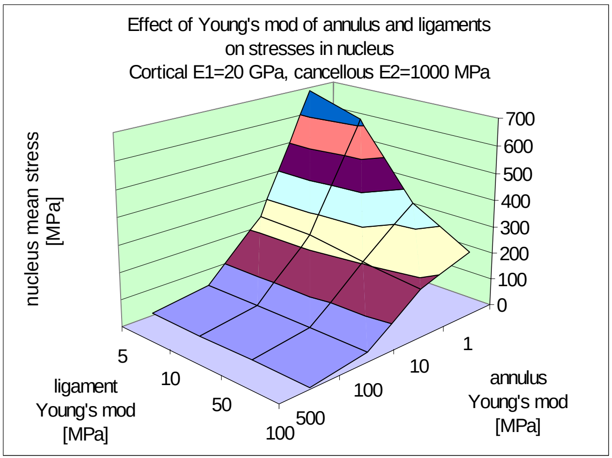


Figure 5: The coupled effect of Young's moduli of disc annulus and ligaments on the stresses of disc nucleus

The results show that the average stresses both in the annulus fibrosus and in nucleus pulposus are almost constant independently of the Young's modulus of the ligaments, if the annulus fibrosus has high tensile modulus of elasticity. When it is smaller than 10 Mpa, the average stresses both in the annulus fibrosus and in the nucleus pulposus decrease by increasing the Young moduli of the ligaments

The parameter identification aims to determine the in vivo controlled tensile behaviour and tensile material moduli of the component organs of lumbar segments that are completely missing in the international literature. Namely, except for the results of *Kurutz et al.* [1,2] there can hardly be found any experimental results for in vivo human lumbar spine in pure centric tension, when the effect of muscles are excluded. In parameter identification process, the in vivo measured global elongations of lumbar segment complex are used for control parameters in determining the material moduli of the local component organs.

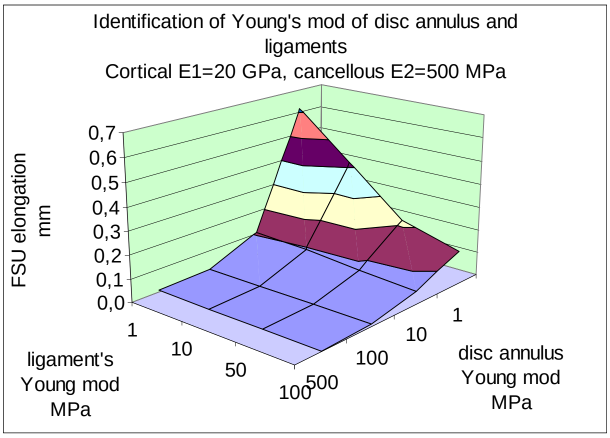


Figure 6: Identification of the Young's modulus of disc annulus and ligaments, by means of in vivo measured global elongations of lumbar spine segments

In this process, the material moduli of certain organs are kept constant while the material parameters of other organs are considered to be the variable of the problem. Under these parameters, numerical simulation is investigated, and those results are considered to be realistic and are accepted that leads to the in vivo measured elongations.

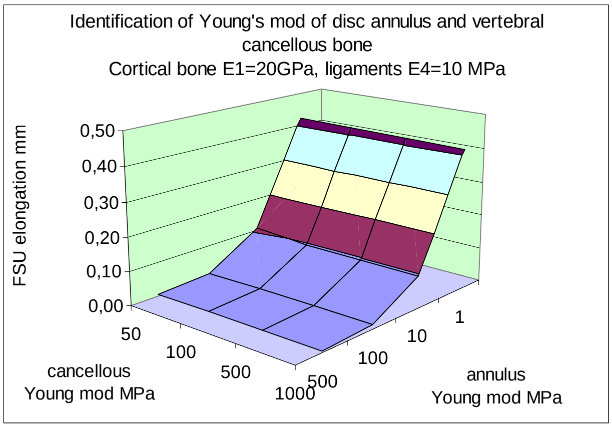


Figure 7: Identification of the Young's modulus of disc annulus and cancellous bone by means of in vivo measured global elongations of lumbar spine segments

In Fig. 6. and 7. the global elongation of the segment complex is illustrated in terms of the Young's modulus of the disc annulus and the ligaments, and the disc annulus and the cancellous bone of vertebrae, respectively. By considering the in vivo measured global elongation, for example, as 0,3 mm, the intersection line of the horizontal plane of this elongation level and the surface can be projected to the basic coordinate plane, and so the possible coincidence function of the of the real Young moduli of the above mentioned organs can be obtained.

Here we can observe again that the Young's modulus of the annulus has a significant effect to the deformability of segment complex, while the vertebral bone has a very small effect on the tensile deformability of FSU.

## Conclusions

Based on in vivo measured elongations of human lumbar spine segments, numerical creep models of human lumbar-lumbosacral segments L3-S1, elastic and damping moduli have been obtained for pure centric tension. By means of the initial elastic moduli and the damping parameters, it has been numerically shown that the rigidity of segments increases significantly with aging. Moreover, the time related difference between the deformation propagation of male and female patients depends on the different damping properties of sexes. The values of elastic and creep parameters are approximate, since they are very sensitive to the measured deformations. However, the ratio and tendency of them provide very important information for the numerical simulation of traction therapies. Numerical simulation and parameter-identification of human lumbar spine segments was presented in centric tension. 2D and 3D FEM models were used, and the global elongations of FSUs were used for identification of the local material


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parameters of the component organs of lumbar spine segments.

Except for the results of *Kurutz et al.* [1,2,3,4] there are no in vivo experimental results in the international literature for the human lumbar spine segments or discs in pure centric tension when the effect of muscles can be excluded. Thus, the presented results may arouse the interests of the biomechanical analysts of the human lumbar spine.

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