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In vivo deformability of human lumbar spine segments in pure centric tension measured during traction bath therapy

M. KURUTZ, A. LOVAS

Budapest University of Technology and Economics, Budapest, Hungary

E. BENE

Hungarian Institute of Rheumatology and Physiotherapy, Budapest, Hungary

In vivo tensile deformability of human lumbar spine segments has been measured during the usual traction bath hydrotherapy of patients when the effect of muscles can be excluded, and the lumbar spine is subjected to pure centric tensile load. Elongations of segments LIII-IV, LIV-V and LV-SI have been measured by using a special computerised subaqueous ultrasound measuring method developed by the authors. Influence of biomechanical parameters, aging, sex, body weight and height, body mass index (BMI) has been established; moreover, time-dependence of deformability has also been measured. The aim of the experiments was double: partly to clear the tensile effect of the traction bath therapy, and partly to obtain the in vivo numerical traction model of human lumbar motion segments. This report deals with the in vivo tensile deformability of lumbar segments only. The biomechanical parameter analysis and the time-dependent numerical creep model of human lumbar spine segments will be reported in forthcoming papers.

Key words: tensile deformability, traction bath hydrotherapy, lumber spine segments, numerical traction model

1. Introduction

As far as we know, in vivo experimental results for deformability of human lumbar motion segments subjected to pure centric tension, when the effect of muscles can practically be excluded, have not been documented yet. Biomechanical experiments of lumbar spine are related generally to physiologic loading cases: compression, flexion, extension, lateral bending or torsion, by using cadaveric motion segments as test specimens [4]. Moreover, tensile stresses analyzed are associated with flexion and extension. In this paper, in vivo elongations of human lumbar spine segments being presented are measured during the usual traction bath hydrotherapy of patients. In such a case, the motion segments are subjected to pure centric tension.

The so-called “weight-bath” therapy is an original Hungarian invention introduced by MOLL [20] in order to stretch the spine and the lower limbs. In traction bath hydrotherapy, patients are suspended cervically or supported by armpit bars in lukewarm water for 20 minutes, with or without extra lead weight loads applied to the ankles or to the waist. Since the traction bath hydrotherapy is applied in Hungary in the everyday clinical practice for half a century without any biomechanical control, the first aim of experimental research was to document the time- and load-related spinal deformations occurring during the treatment. In addition, the 20 min treatment process came in very handy for investigating a biomechanical parameter analysis to find the time-, aging-, sex-, body weight- and height-dependent numerical models of lower lumbar motion segments in pure centric tension, without the effect of muscles. The biomechanical traction model of human lumbar motion segments concerns the numerical simulations of traction therapies to clear the mechanical behaviour and role of different organs composing a very complicated biomechanical structure of spinal motion segments.

More than 3000 ultrasound pictures of about 400 lumbar segments of 155 patients have been measured and evaluated. Of a huge amount of biomechanical results of the 3 year large-scale experimental analysis, in this paper the in vivo tensile deformability of lower lumbar segments is reported only. The biomechanical parameter analysis and the time-dependent numerical creep model of human lumbar spine segments as well as the results of the numerical simulations will be reported in forthcoming papers.

2. Methods

The experiments have been carried out in the Hungarian Institute of Rheumatology and Physiotherapy during the usual prescribed traction treatment of patients. A competent ethical committee gave permission for experimental investigation with human subjects. The patients properly informed about the experiment also gave their consent.

Deformations of segments have been measured during the usual traction bath therapy of patients, whose different parts of spinal column or lower limbs were subjected to stretching. The traction bath therapy itself has been introduced and described by MOLL [20]–[23], and later by BENE [5], while the biomechanics, indications and contraindications of the therapy have been detailed by BENE and KURUTZ [6]. The therapy needs a specially deepened basin with elastic suspension equipment. During the traction bath treatment, the patients are suspended cervically or by armpit supports in lukewarm water for 20 minutes, loaded with extra lead weights applied to the ankles or to the waist. A collar guarantees a natural and comfortable position of the head of patient in the state of cervical suspension (figure 1).

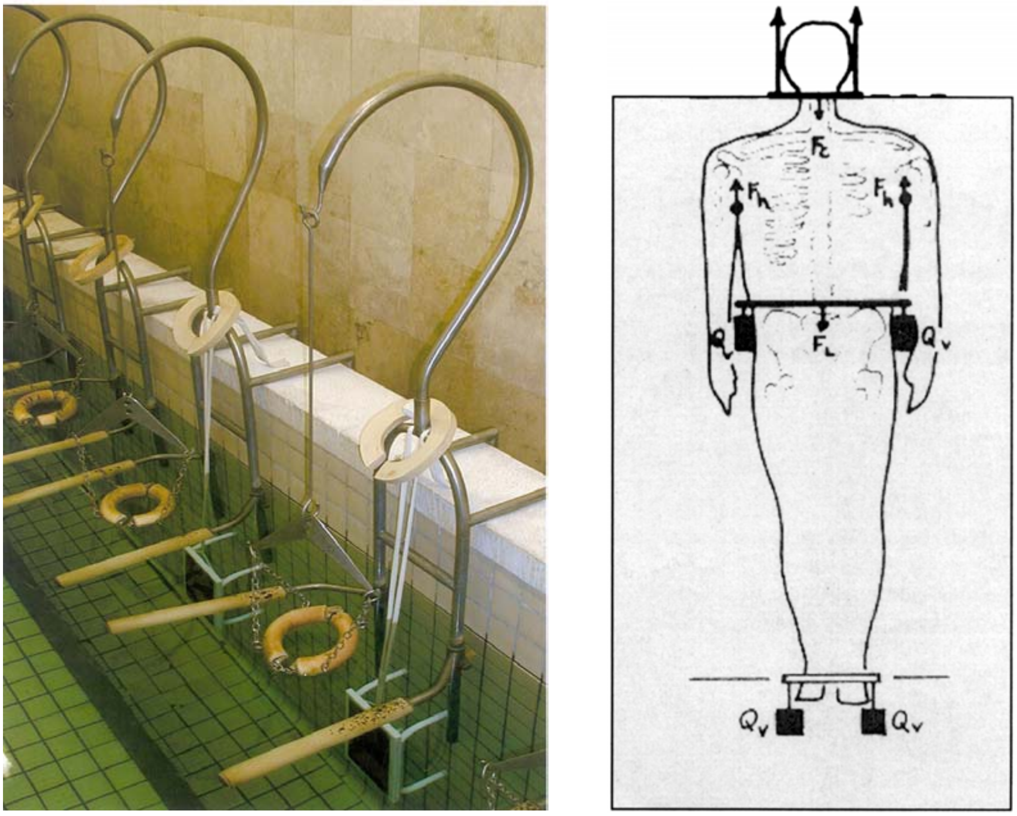

Fig. 1. The traction bath suspension equipment and treatment

During 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. The extra lead loads have been applied to *ankles* only.

2.1. Traction forces in weight-bath treatment

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. 3. The *extra loads* applied in the therapy. In the case of free cervical suspension in water, the effect of muscles can be neglected. The distribution of these traction load effects is illustrated in figure 2, beside the suspended body, where the load distribution diagram can be seen. The compressive loads of upright standing position are seen to the left to the vertical zero axis, while the active tensile loads of suspension are illustrated to the right.

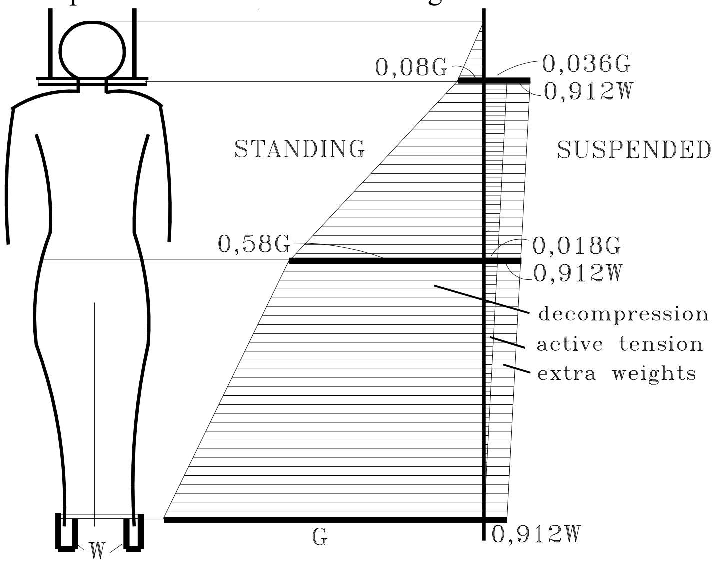

Fig. 2. Compressive and tensile loads of spine in standing and suspended position

1. In normal upright standing position, the *decompressive force* is equal to the compressive load acting on the spine before the treatment. This gravitation force increases continuously downwards, starting from zero, namely from the top of the head to the floor, where the total weight G of the body acts (figure 2). At the cervical level this force is equal to the weight of the head, that is, about 8% of the body weight. At the lumbar level the decompressive force takes about 58–60% of the body weight, namely, the weight of the head and trunk together. That is:

$$F_1 = 0.58G.$$

For example, in the case of cervical suspension of a patient whose weight G is 700 N, the decompressive force at the lumbar level is about $F_1 = 0.58\ G = 406$ N. In the case of armpit supports, the weight of the head (0.08 G) acts on the spine further on, thus the decompressive force is smaller. Moreover, by using armpit supports, the contracting effect of muscles that decreases the efficiency of traction treatment needs to be considered as well.

2. The *active tensile force* and the extra loads have been calculated by BENE and KURUTZ [6]. These forces depend on the buoyancy. The active tensile force is due to the difference between the body weight and the buoyant force. It depends on the difference between the density of the body and the density of water as well as on the distance from the point of suspension. This force decreases continuously along the

spinal column, starting from the suspending point where it reaches the maximum value (figure 2). According to [6], when a patient is suspended cervically in water without any artificial weights, the value of the active tensile force acting at the suspension point is about $0.92\,G(1-\rho_w/\rho_b)$, where ρ_w and ρ_b are the densities of water and human body, respectively. At the lumbar level approximately the half of this force is valid, namely, the active tensile force can be considered as

$$F_2 = 0.46G(1-\rho_w/\rho_b)\,.$$

Considering an average density of human body $\rho_b = 1040\ \mathrm{kg/m^3}$ and a normal water density $\rho_w = 1000\ \mathrm{kg/m^3}$ yields

$$F_2 = 0.46G(1-\rho_w/\rho_b) = 0.018G\,.$$

For example, in the case of a patient whose body weight $G = 700\ \mathrm{N}$, the lumbar active tensile force yields only $F_2 = 0.018 \cdot 700 = 12.6\ \mathrm{N}$. Moreover, this effect is even smaller in thermal water, the density of which is higher than that of normal one. Thus, the active tensile load seems to be surprisingly low.

3. The *extra weight loads* are applied to provoke a more intensive stretching effect in the lumbar spine. The extra weight load depends on the value, specific weight and the place where they are applied (compare also with [6]). In our experiments, the extra loads have been applied exclusively at the ankles of patients (figure 2). Since the extra lead weights are also in the water, the original value W of them is reduced to

$$F_3 = W\,(1-\rho_w/\rho_l)$$

due to the buoyancy. By considering the density of the lead as $\rho_l = 11350\ \mathrm{kg/m^3}$, we arrive at the extra weight load

$$F_3 = W\,(1-\rho_w/\rho_l) = 0.912W\,,$$

causing a constant load value along the spinal column (figure 2). Thus, by substituting the applied actual weights of $2 \cdot 20$ N, that is $W = 40$ N, we obtain $F_3 = 0.912 \cdot 40 = 36.48$ N.

During the traction bath treatment, the sum of the detailed three tensile forces is to be considered:

$$F = F_1 + F_2 + F_3 = 0.58G + 0.018G + 0.912W = 0.598G + 0.912W,$$

by supposing the above-considered densities. By comparing the three component forces, obviously, the dominant stretching load is the decompressive force, provided that cervical suspension is applied. It takes about the 97% of the stretching load if no extra loads are applied. In the case of armpit bar supports, the forces and the action domain of them are different, the decompressive load is smaller; moreover, the effect of muscles have to be taken into account as well, thus, the stretching load is much smaller and the lumbar traction is less effective. *Even the mode of cervical suspension provides the optimal stretching effect in the lumbar spine.* 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 and allow the stretching effect to develop optimally.

It is worth noting here that by considering the dominant stretching effect of decompression, the importance of simple *swimming* as traction therapy of low back problems can be verified. Indeed, the decompression occurs due to swimming, that is, the 97% of the stretching load work, being very effective in unloading the spine. However, by swimming, namely by keeping the body on a water surface, the muscles actively contract, thus the stretching effect is less effective than in inactive relaxed position of the cervical suspension of traction hydrotherapy.

Consequently, *in our experiments, cervical suspension has been applied exclusively*. Naturally, the measurements were applied to those patients only, to whom the cervical suspension was prescribed by medical indication. Those patients, to whom the cervical suspension was contraindicated and other modes of support were ordered medically, have been excluded from the experiments.

2.2. The subaqueous ultrasound measuring method

Tensile deformations of the segments suspended in water are, *by definition*, *considered as the decrease of compressive deformations existing before the treatment*. Namely, zero deformation belongs to a common compressed state of segments of normal upright position of body. *Tensile deformation of segments has been considered as the change of the distance between two spinous processes of neighbouring vertebrae*. To avoid the rotational effects of segments, a rigid plate was mounted on the wall of the basin to keep the spine segments in parallel position during the deformation process. Elongation of segments has been specified to be the measured deformation compared with the state of the segments just before the traction bath treatment.

The *elongations have been measured by using a special subaqueous ultrasound measuring method developed by the authors* (figure 3). The weight-bath set has been equipped with a mobile ultrasound instrument connected to a computer. Hitachi EUB 405 mobile instrument with a transducer of the 7.5 MHz frequency has been used. Common personal computers were applied in both the hospital and research center. Mobile winch has been used for transporting the measurements of experiments. Storing, treatment, evaluation of results and archiving of the documents have been arranged at the research center.

The cervical suspension has been applied for the regular 20-minute treatment time. Elongation of lumbar segments LIII-IV, LIV-V and LV-SI has been measured. The ultrasound pictures and measurements results have been computerized, digitally stored and evaluated by using a method for analyzing the pictures developed by the authors. More than 400 lumbar segments of 155 patients have been measured, more than 3000 ultrasound pictures have been evaluated. Based on the in situ measurements used as reference values, the ultrasound pictures have been compared with each other to demonstrate the deformation process of segments.

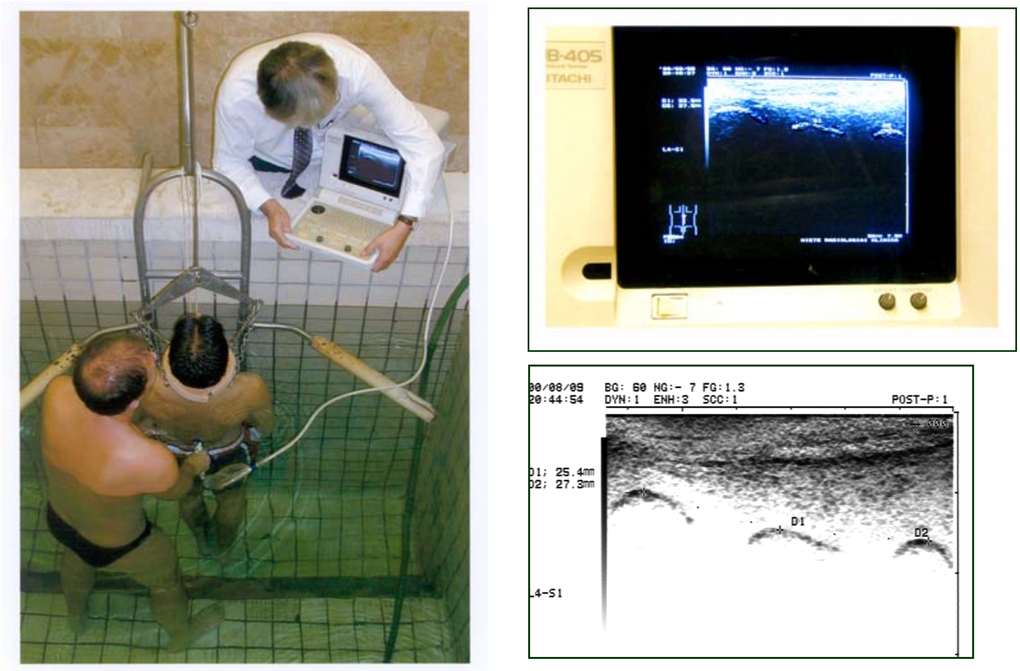

Fig. 3. The subaqueous ultrasound measuring method

A total of 155 patients have been divided into two groups: 88 patients with extra lead loads, that is, 20–20 N applied symmetrically to the ankles, and a group of 67 patients 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 discs. Table 1 illustrates the age and sex distribution of the patients examined.

Table 1. Age and sex distribution of patients

Age and sex distribution	Total of 155 patients			With extra weights, 88 patients			Without extra weights, 67 patients		
Group of ages	Male	Female	Total	Male	Female	Total	Male	Female	Total
Under 20 years	2	–	2	1	–	1	1	–	1
20–30	10	4	14	4	3	7	6	1	7
31–40	4	5	9	3	2	5	1	3	4
41–50	18	28	46	13	15	28	5	13	18

51–60	21	26	47	14	18	32	7	8	15
61–70	8	15	23	3	8	11	5	7	12
Over 70 years	4	10	14	2	2	4	2	8	10
Total	67	88	155	40	48	88	27	40	67

In the case of group with extra weights, deformations have been measured in four phases of the treatment: just before the treatment; immediately after being suspended in the water without extra weights (t = 0 min); immediately after the preceding measurements with extra weights (t = 3 min); at the end of the 20-minute suspension (t = 20 min). In some cases, measurements have been carried out after the therapy, just by leaving the basin, as well. For the group with no extra weights, three phases of measuring have been distinguished: just before the treatment; immediately after being suspended in the water (t = 0 min); at the end of the 20-minute suspension (t = 20 min).

Due to certain degeneration of segments, the respective ultrasound pictures could not be clearly recognized and evaluated. In the group of 88 patients with extra weights, 7 (8%) of segments LIII-IV, 2 (2%) of segments LIV-V, and 19 (22%) of segments LV-SI were unmeasurable. In the group of 67 patients without extra weights, 3 (4%) of segments LIII-IV, 2 (3%) of segments LIV-V, and 23 (34%) of segments LV-SI could not be measured nor evaluated.

By using the method developed, segment deformations less than 0.2 mm cannot be observed and registered; consequently, deformations within this range have been considered to be zero. Segments having non-observable small deformations have been considered to be undeformed.

3. Results

Consider first the change of *ratio* and later the change of *value* of in vivo tensile deformability of human lumbar spine segments during the time of treatment.

3.1. Change of *ratio* of tensile deformability during the traction treatment

Table 2 illustrates the percentage of deformability of segments for groups with and without extra weights, in different scale of elongation values, for the phases of treatment, for segments LIII-IV, LIV-V and LV-SI, respectively, by distinguishing the sexes. Percentage of undeformed and deformed segments is indicated in the first and last shadowed column of table 2.

In the group with extra weights, for segments LIII-IV, the ratio of their deformation was 52% at suspension (t = 0 min), 74% at suspension with extra weights
(t = 3 min), and 85% at the end of the treatment (t = 20 min), thus finally 15%

remained undeformable. For segments LIV-V, the deformability increased from 47% through 60% to 70%, and 30% remained undeformable, while for segments LV-SI, it changed from 29% through 49% to 68%, and 32% remained undeformable. The ratio of deformability was lower in distal direction, while, parallelly, the rigidity was higher.

In the case of group with no extra weights, the results were similar, but deformability proved to be lower. For segments LIII-IV, 34% extended at suspension and 64% at the end of the treatment, and 36% remained undeformable, for segments LIV-V the ratio increased from 37% to 60%, and 40% remained undeformable, for segments LV-SI, it changed from 34% to 55%, while 45% remained undeformable. The ratio of undeformed segments was higher in distal direction in this group as well.

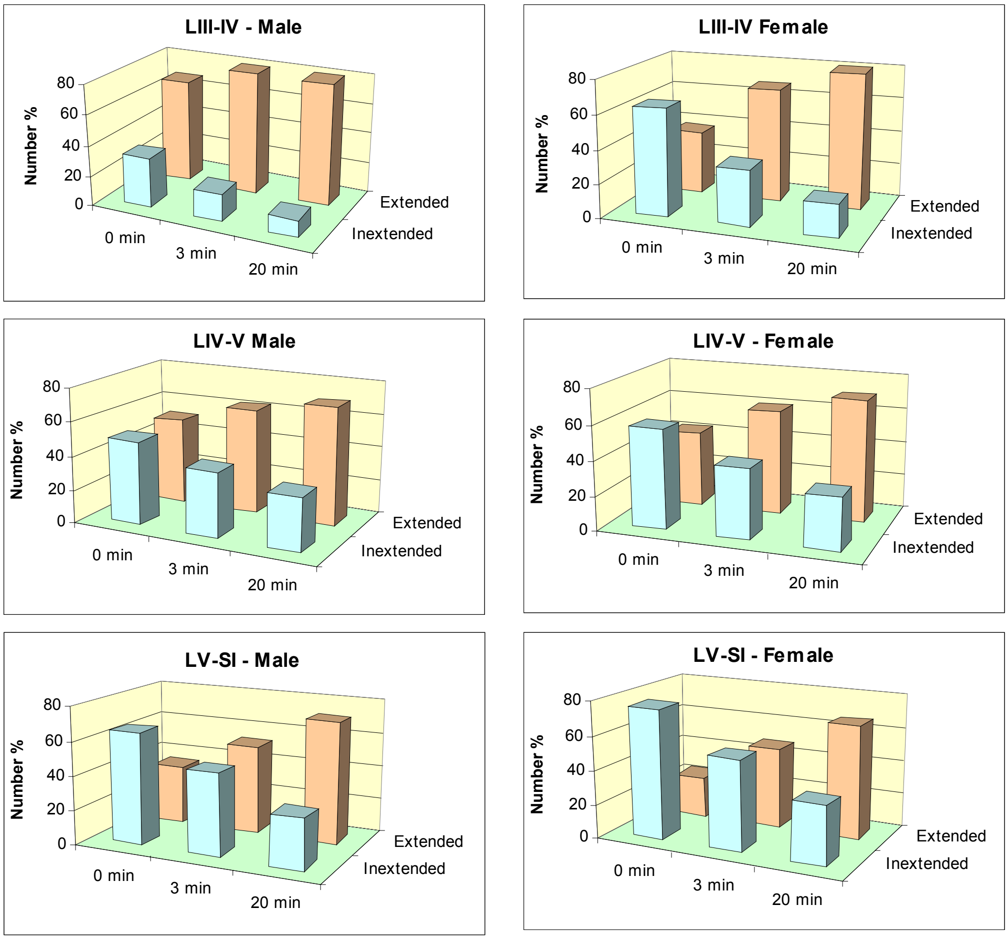

Fig. 4. Ratio of deformability of segments versus sex in the group with extra weights

In figure 4, the percentages of extended and inextended segments are distinguished for segments LIII-IV, LIV-V and LV-SI, for men and women, at three time instants of the treatment, for the group with extra weights. Deformability ratios for male and female patients can be compared. Significant difference has been observed between men and women: at the beginning, the ratio of extended segments is higher in men, while at the end of the treatment, this ratio in women practically approaches that of men.

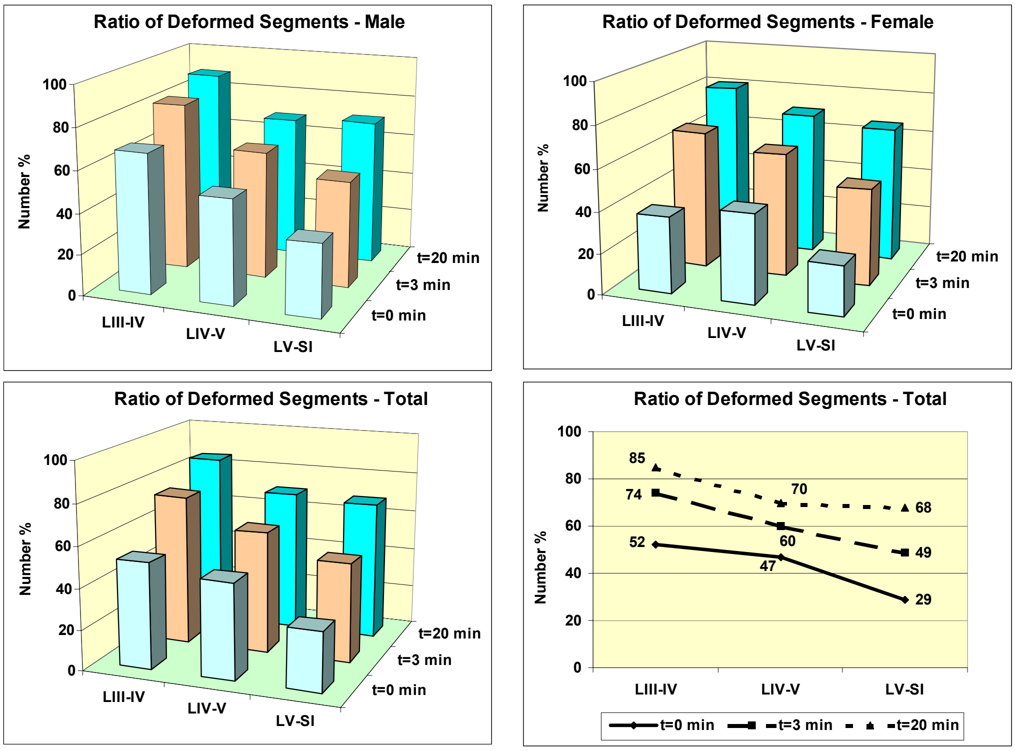

Fig. 5. Ratio of deformed segments versus position of segments in the group with extra weights

In figure 5, the time-related change of deformability ratio is demonstrated in terms of the position of segments for the group with extra weights, by distinguishing the sexes. The observation that the deformability of segments decreases in distal direction in both male and female patients in any time of the treatment has been verified. This decreasing tendency is illustrated in the last diagram of figure 5.

3.2. Change of the *values* of tensile deformability during the traction treatment

Beside the ratio of deformed to undeformed segments, in table 2 the scales of elongations are illustrated as well. Table 3 contains mean values of elongations and tensile strains of segments for several phases of the traction hydrotherapy for both groups. Both mean elongation and strain values calculated are averaged for all (deformed and undeformed) segments. Since deformations less than the 0.2 mm observation limit have been considered zero, mean values in table 3 are slighty lower than real mean values. Thus, the values in table 3 can be considered as the lower bound of the real mean elongations and strains of segments. The strains, namely the dimensionless deformations, are obtained by dividing the measured deformations by

the heights of segments. The segment heights have been calculated as a function of the body height and the position of the segment. This calculation supposes the strain to be constant along the height of the segment.

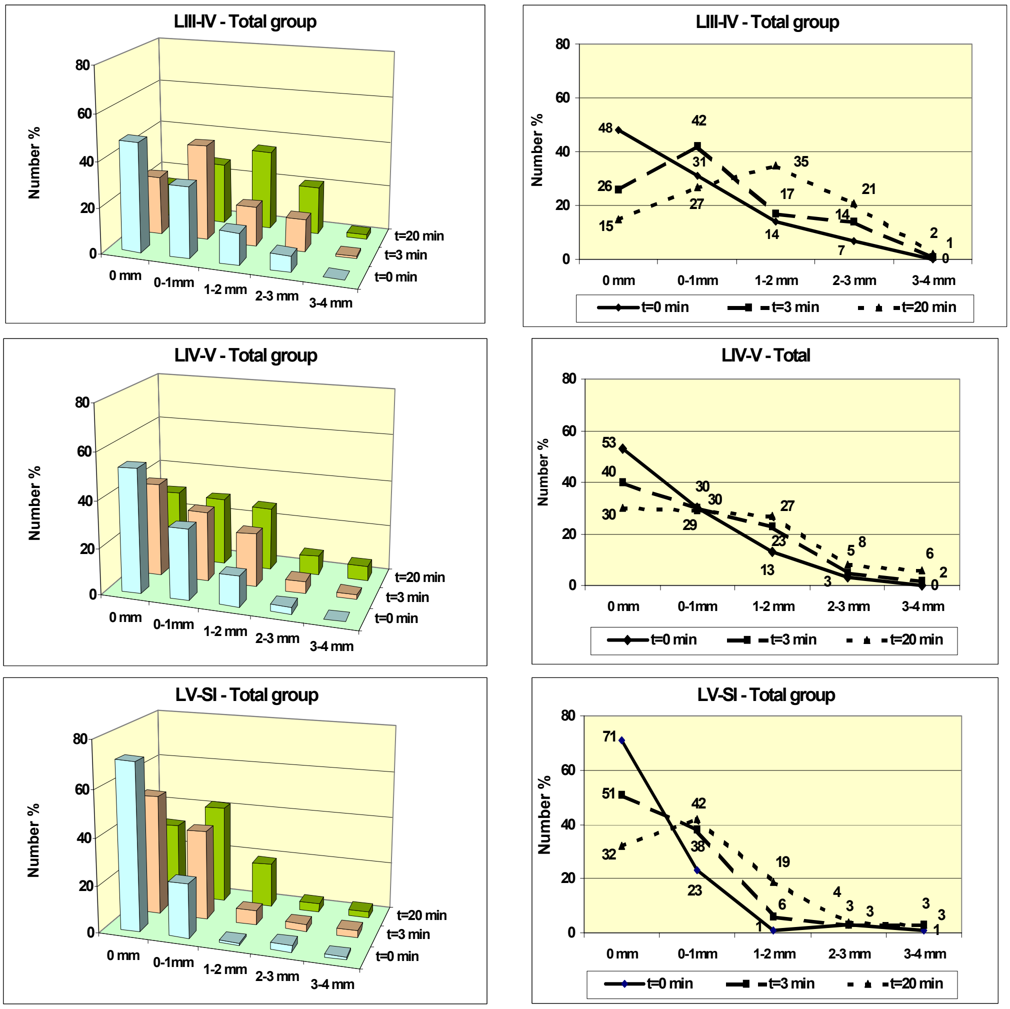

Fig. 6. Scale and distribution of deformability of segments versus time
in the group with extra weights

At the beginning, just at that moment the patients are being suspended in the water (t = 0 min), 30–50% of them in both groups show instantaneous extension in lumbar segments (table 2). Just at suspension due to a sudden and rapid decompression, i.e. the removal of compression and the buoyancy, unloading of discs and segments occurs, thus, even without any extra weights a significant tensile extension has been registered. These elongations correspond to elastic behaviour of segments with a mean value of 0.4–0.6 mm instantaneous elongation (table 3). Note that significant difference can be observed in the initial deformability of men and women: for men about 0.4–0.8 mm, for women about 0.3–0.4 mm mean instantaneous elastic deformation occurs.

Being suspended in the water with extra weights of 20–20 N on the ankles (t = 3 min), 50–75% of patients show extension (table 2), and a mean value of 0.6–1.0 mm elongation has been registered per segment (table 3).

At the end of the treatment (t = 20 min), elongation have been demonstrated practically in 55–65% of patients in the group with no extra weights, and in 70–85% of patients whose ankles were loaded with 20–20 N (table 2). A total mean elongation of a lower lumbar segment after a 20-minute traction bath treatment with cervical suspension without any extra weight loads is about 0.8–0.9 mm, while with 20–20 N extra weights at the ankles it is about 0.8–1.4 mm (table 3). However, in spite of the initial difference in the deformability of men and women, final deformations are practically the same.

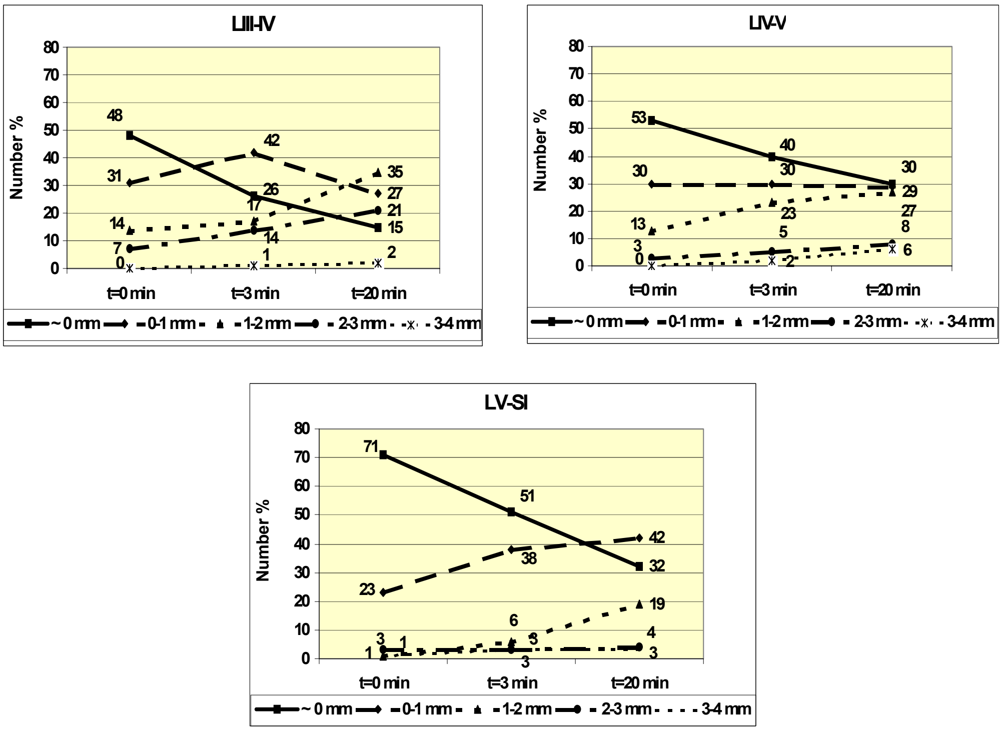

Fig. 7. Change of deformations of segments versus time in the group with extra weights

It has been observed that the 20-minute treatment time in itself has a significant traction effect. Mean additional elongation without extra weight loads is about 0.4–0.5 mm per segment, while with 20–20 N extra weights it yields about 0.5–0.8 mm per segment. Thus, the traction effect of 20–20 N extra weights during 20 minutes can be considered as 0.2–0.3 mm elongation per segment.

Figures 6 and 7 show the percentage of segments having different scales of elongation values for three phases of treatment of the group with extra weights for segments LIII-IV, LIV-V and LV-SI. The change of ratio of the percentage of rigid segments (0 mm) to the segments with 0–1, 1–2, 2–3 and 3–4 mm extensions is illustrated in time. It can be seen that both the ratio and the value of tensile deformations increase during the treatment time for all three segments.

According to figure 6, where the distribution of deformability values is presented during the 20-minute treatment time, the number of rigid segments decreases, and the number of segments with larger elongations increases which can be seen in figure 7 as well. At the beginning (t = 0 min), a monotonically decreasing function characterizes the distribution of deformability in figure 6. At the end of the treatment the function tends to the normal distribution function of Gauss: equally few segments with no deformations and with extra large (3–4 mm) deformations, and most of the segments with the deformations in the range of 1–1.5 mm. Naturally, the medically limited 20-minute treatment stops the function to produce the perfect form of the Gauss curve, since longer treatments may be invasive and dangerous for the patients.

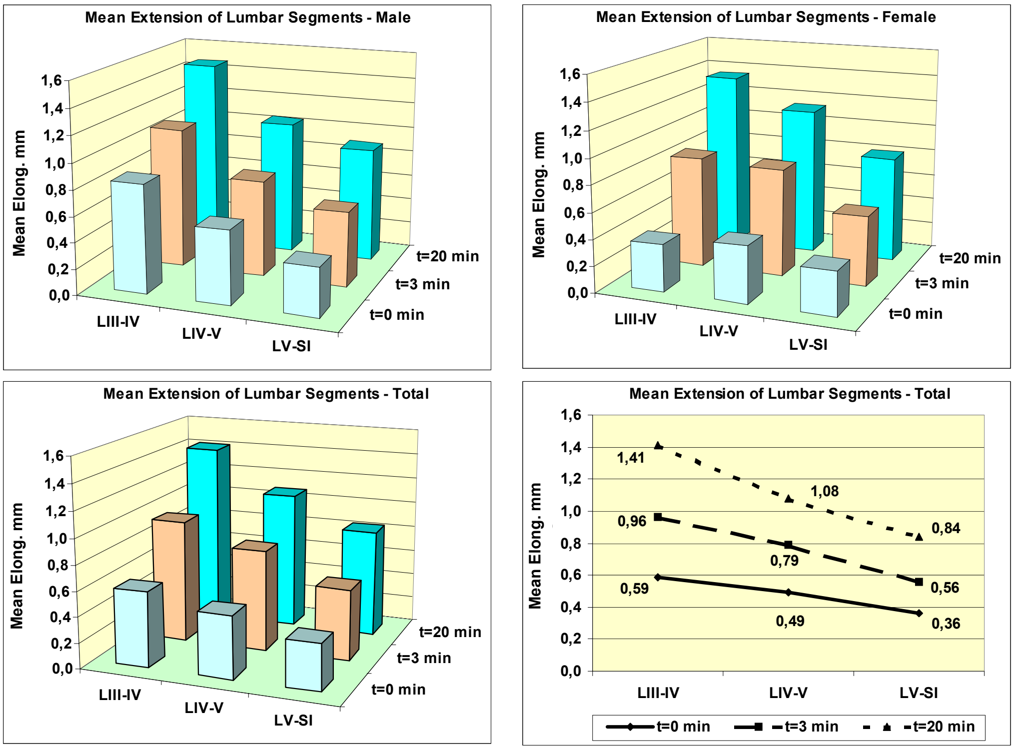

Fig. 8. Mean extensions versus position of segments in the group with extra weights

Figure 8 illustrates mean elongations of lumbar segments in terms of the position of segments for the group with extra weights, by distinguishing the sexes. It has been numerically verified that the rigidity of segments is higher in distal direction, i.e., the deformability decreases in distal direction. As can be seen in the last diagram, the trend of the deformability decrease in distal direction is more significant at the end of the treatment.

3.3. Effect of biomechanical parameters on tensile deformability

During the time- and load-related traction treatment, the effect of sex, body height and body weight on tensile deformability of lumbar segments has been registered.

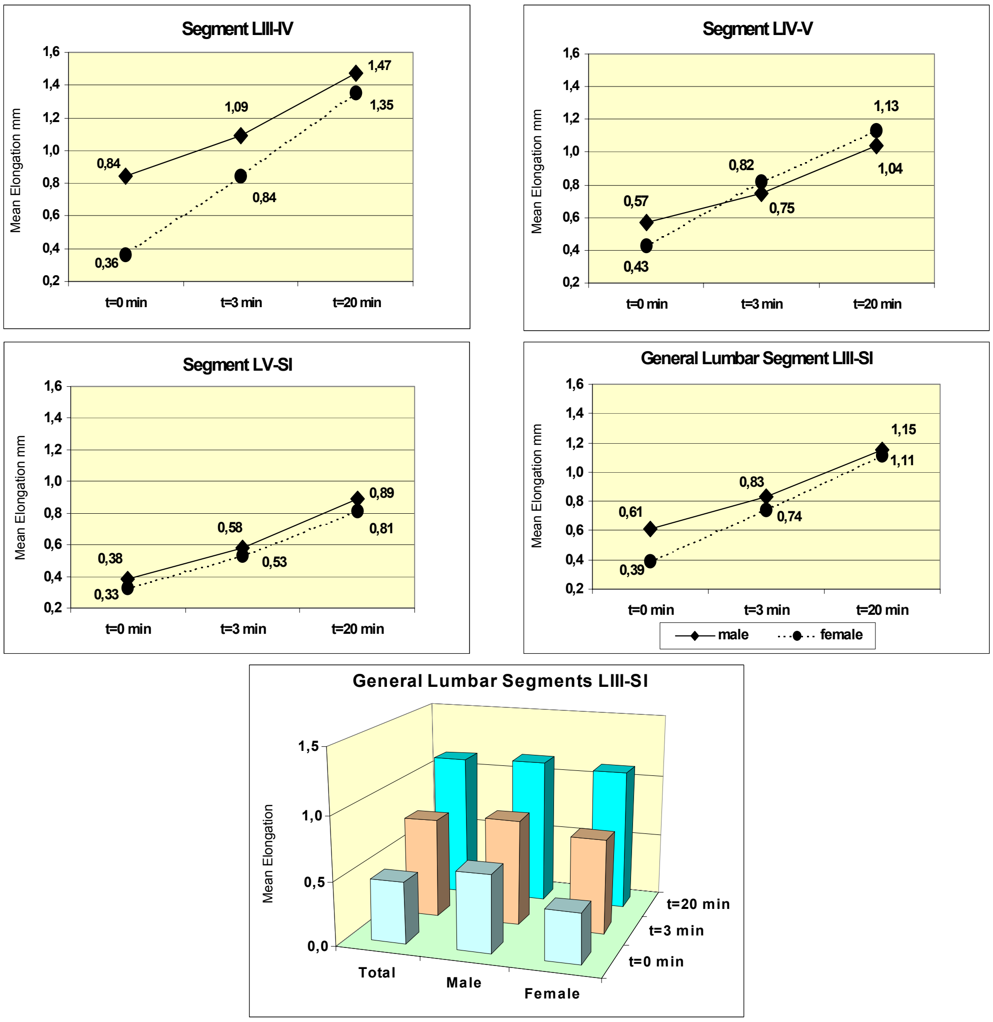

Fig. 9. Tensile deformability of segments versus sex in the group with extra weights

The *effect of sex* can be compared in figures 4 and 9, where the ratio the value of tensile deformability of segments, respectively, are illustrated. In figure 9, the time-related mean elongations in men and women are compared for each measured segment and for a general lower lumbar segment LIII-SI. A significant difference in reaction time is revealed depending on a patient's sex. Women react more slowly to the traction treatment, which means that more female segments remain undeformed at t = 0 min compared to male segments. However, at the end of the 20-minute treatment, the extension effect in women seems to very close to that in men. At the end of the treatment not only the ratio of extended segments but also the value of extension seem to be equal for both men and women. Therefore can conclude that the deformation propagation occurs in a different way for sexes, but with the same final result.

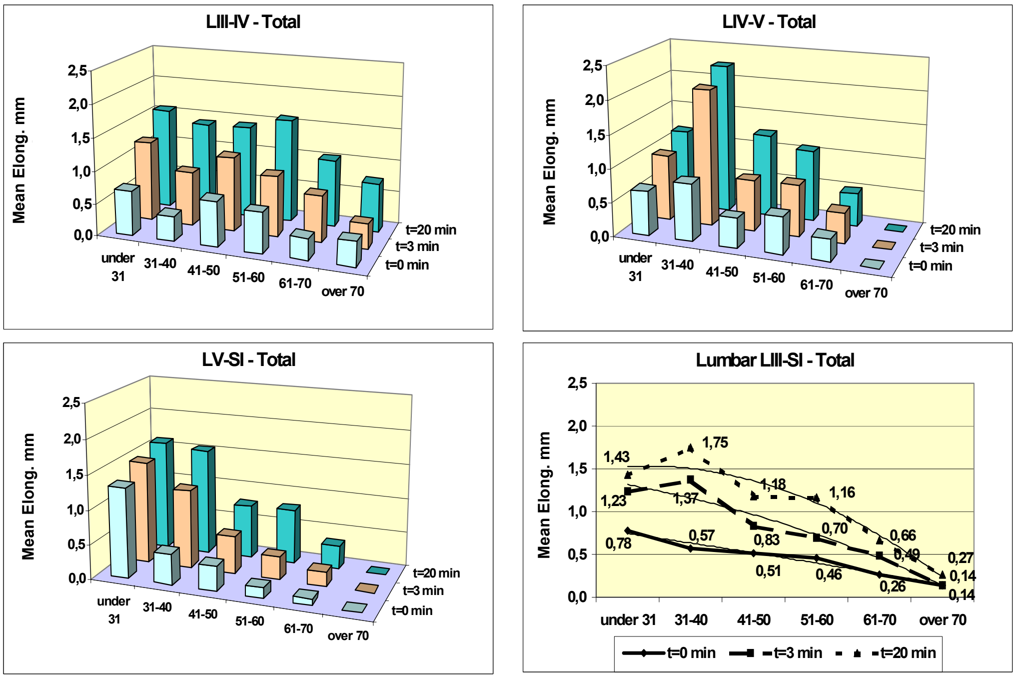

Fig. 10. Tensile deformability of segments versus aging in the group with extra weights

The *effect of aging* can be assessed based on table 4 and figure 10. According to the parameter analysis, it has been numerically verified that the deformation capacity of lumbar segments is sensitive to aging. In table 4, mean ages, mean body weights as well as heights of the patients with deformed and undeformed segments are compared for the group with extra weights, by distinguishing the sex. According to table 4, the patients with deformable segments are significantly younger than the patients with undeformable segments. While in the case of the segments LIII-IV and LIV-V, deformable patients are, respectively, only 3–6 and 6–9 years younger, in the case of

segments LV-SI the difference is 10–13 years. In the groups with more rigid segments, deformability occurs even at younger age. Apart from some exceptions, practically there is no significant difference between men and women, which could be due to age. It has been verified experimentally that deformability decreases with increasing age of patients in any time of treatment (compare figure 10 for each segment). By considering the age-dependence of each segment in figure 10 we arrive at conclusion that the age effect seems to be more significant in distal direction. In figure 10, the trend lines of the age-sensitivity are also shown for a general lumbar segment LIII-SI. Instantaneous elastic state (t = 0 min) can be represented by linear relation, while for the time-dependent further states a higher-order approximation seems to be acceptable. By using linear approximation for each phase of treatment, we have found that the slope of trend lines increases in time, thus, the age-sensitivity seems to be increasing in time.

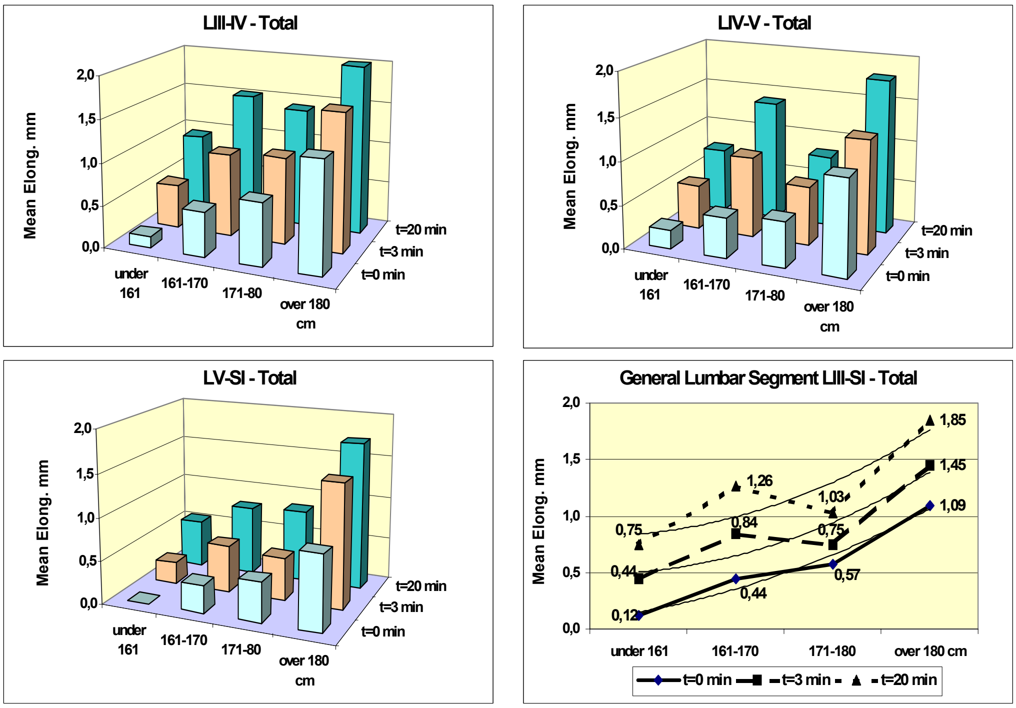

Fig. 11. Tensile deformability of segments versus body height in the group with extra weights

The *effect of body height* is presented in table 4 and in figure 11 for the group with extra weights. It has been verified that the deformation capacity of lumbar segments is proportional to the body height: the deformations increase with increasing height of patients (table 4). Immediately after being suspended the patients with deformed segments were by about 4–6 cm taller than the patients with inextended segments, and by about 1–5 cm taller at the end of the 20-minute treatment. In figure 11, where the second-order approximation has been supposed, the trend lines of height-dependence are presented. The height-sensitivity of segments seems to be similar in any time of the treatment. However, in the group with no extra weights, the effect of body height could be registered at the beginning of the deformation process only (1–4 cm), while at its end an inverse tendency has been found. Taking account of the constitutional differences in body heights of men and women, there seems to be no significant difference between them regarding height effect.

Taking account of the *effect of body weight*, no unambiguous correlation has been found between the deformability of lumbar segments and the weight of the patient. A low correlation has been observed only in the initial elastic phase for the group with extra weights (table 4 and figure 12). At the beginning of the loading process, deformation capacity increased slightly with increasing body height. However, further on and 20 minutes later, this tendency was not valid any more. Moreover, for the group with no extra weights, due to more degenerated segments, there was no correlation between deformation capacity and body weight at all.

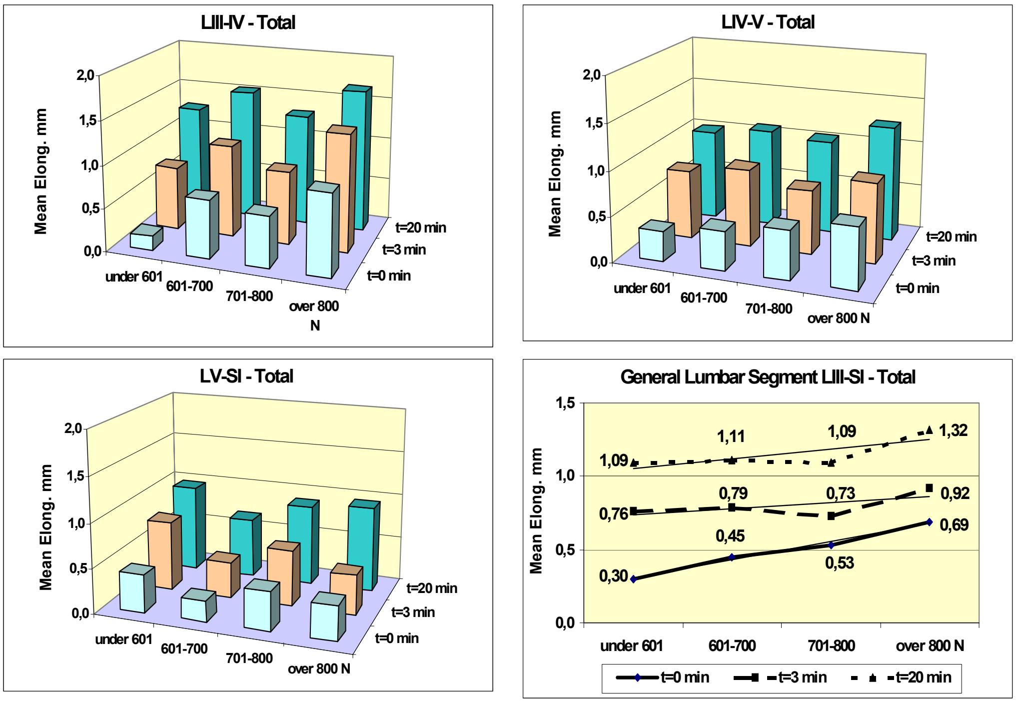

Fig. 12. Tensile deformability of segments versus body weight in the group with extra weights

4. Discussion and conclusions

Based on the computerized underwater ultrasound measuring method developed by the authors and applied to about 400 lumbar segments of 155 patients during their regular traction bath treatment, regarding the tensile deformability of lumbar segments LIII-IV, LIV-V and LV-SI, the following observations and conclusions are offered for discussion.

1. The presented *subaqual ultrasound technique* applied during traction bath therapy in order to measure the elongations of lumbar spine segments in vivo is a non-invasive, reproducible method. It allows documenting in vivo tensile deformability of human lumbar motion segments subject to pure centric tensile load with no effect of muscles.

2. Traction deformations of human lumbar spine segments have been measured in vivo during the *traction bath hydrotherapy* usually prescribed for patients having more or less degenerated discs or segments. 88 patients were suspended with extra lead loads (20–20 N at the ankles), and 67 patients without any extra weights. More degenerated segments of the patients in the latter group were a contraindication for extra weights.

3. *Cervical suspension* has been applied for 20-minute treatment. It has been numerically verified by biomechanical calculations that the cervical suspension provides the most effective traction load for the lumbar spine.

4. Weight-bath therapy seems to be appropriate to analyze lumbar traction effects in *pure centric tension without the influence of muscles*. ANDERSSON et al. [3] reported that by active traction in laying posture, the back muscles contract and the disc pressure increases. Similarly, RAMOS and MARTIN [25] verified the inverse relationship between the traction load being applied 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. Indeed, being suspended comfortably in lukewarm water, the body is unloaded due to buoyancy; moreover, the patients nearly sleep, thus, the muscles relax, and the stretching effect can optimally develop.

5. *Deformation of segments* has been specified and measured as the change of the distance between two spinous processes of neighbouring vertebrae. By using non-invasive ultrasound method, of the parts of lumbar segments, spinous processes can be seen only. To avoid the rotational deformations of segments, a rigid plate has been mounted on the wall of a basin to keep the segments in parallel position during the deformation process.

6. *Elongation of segments* has been specified as the decrease of the compressive deformations existing before patient's entering the water. Zero deformation state has been considered as the state of segments before patient's entering the basin.

7. Using the present method, segment deformations less than 0.2 mm cannot be registered. Deformations within this *observation limit* have been considered zero, i.e. the segments with deformations less than those within this observation limit have been considered to be undeformed. Consequently, calculated mean values that include deformed and undeformed segments can be considered as the lower bound, and conversely, calculated mean values that include deformed segments only can be

considered as the upper bound of real mean values. In this paper, mean values are related to all (deformed and undeformed) segments.

8. In spite of the fact that the observation limit is 0.2 mm, the difference in mean values calculated under the same conditions may be smaller than this limit, since *relative results* can be smaller than the observation limit. Evidently, the absolute observation error is equally valid for all results, thus, its effect is eliminated from the difference of certain results. Consequently, the calculated results can be used in biomechanical parameter analyses based on the comparison of certain mean values.

9. During the traction bath treatment consisting in 20-minute cervical suspension in water, the *ratio of deformed segments* increased, and the ratio of undeformed segments decreased. At the end of the treatment a measurable extension is found for 70–85% of segments with extra weight and for 55–65% of segments without extra weights. In the group with extra weights, higher ratio of deformability has been observed. This is partly due to less degenerated segments in this group, and partly due to extra weights.

10. The dominant part of extension occurs immediately at the beginning of suspension. Just after being suspended in water without extra weights, 30–50% of patients show *instant tensile deformation*, with a mean value of 0.4–0.6 mm calculated as an average of all (deformed and undeformed) segments. These deformations are caused mainly by decompression (removal of compression), and by the active tensile force due to buoyancy. Mean elongations due to the unloading of discs correspond to the extension values given between 0.1–1.9 mm per segments also for tension which is reported by BADER and BOUTEN [4].

11. It has been found that the 20-minute treatment in itself has a significant *creep effect*. Mean additional elongations without extra weights and with extra weights were respectively about 0.4–0.5 mm and 0.5–0.8 mm. The observations support the statement that segments and discs have significant viscous characteristics; moreover, on the basis of our experiments, numerical creep model of human lumbar segments has been constructed, which is reported shortly in [11], [14], [16] and in detail in forthcoming papers.

12. As a result of the therapy, at the end of 20-minute treatment, the *total mean tensile deformation* of a lower lumbar segment LIII-SI without extra weight approaches 0.7–0.9 mm, while with extra weights it ranges from 0.8 to 1.4 mm. These results correspond to those reported in the papers dealing with long time, namely diurnal changes of segments and discs. Le BLANC et al. [17] similarly to MALKO et al. [19] found that due to the overnight rest, the decompression of discs yields approximately 1 mm height increment per segment. Diurnal variation in disc height measured by BOOS et al. [7] was 0.85 mm. ROBERTS et al. [26] found that a mean overnight increase in the height of young female's segment was as high as 19.3 mm. According to BOTSFORD et al. [8], this extension is 0.9 mm, while according to LEDSOME et al. [18] an average diurnal change in the lumbar distance LI-LIV reached 5.3 mm. CHEN et al. [9] who described traction therapy found that the intervertebral

disc distance increased by 1.34 mm for prolapsed and by 0.87 mm for normal discs. These results can be used for rough comparison only, for analyzing the effect of decompression without any extra load. On the other hand, the above detailed deformations in the traction therapy occur in a short time, while the diurnal deformations develop during a longer time.

13. Based on the measured elongation of the segments *dimensionless tensile strain* values have been calculated in a usual way: dividing the elongations by the height of the segments. However, these calculated strain values are supposed to be constant along the height of segments, i.e. the segments are far from their real behaviour, thus, they are unusable for the numerical model. Namely, the motion segment is a very complicated structure consisting of two vertebrae with the disc between them, surrounded by anterior and posterior longitudinal ligaments, unco-vertebral joints and muscles. The connections between two contact organs are different, some elements are attached to the other, some elements are free, thus, tensile strains are strongly variable along the height of segments. Numerical simulations are even aimed at clearing the real distribution of strains, thus for the numerical segment models, instead of the calculated constant strains, the measured elongation values are used.

14. It has been numerically verified that both the ratio and the value of deformability of segments are lower in *distal direction*. In the group with extra weights, at the beginning of treatment the mean elongations are 0.6, 0.5 and 0.4 mm, while at the end of the treatment they reach 1.4, 1.1 and 0.8 mm for segments LIII-IV, LIV-V and LV-SI, respectively. Thus, the increasing rigidity of segments has been proven in distal direction. In the group with more degenerated segments, this tendency could not be analyzed because in such a case an extra weight was a contraindication.

15. Significant difference has been demonstrated in deformation propagation between *males and females*: at the beginning women show smaller deformability; however, after 20 minutes the extension effect seems to be almost the same as that in men. This observation proves that though the final tensile deformation capacities of men and women are nearly the same, the way of deformation propagation is different in time. Since there has been observed a slight relationship between the deformability and the weight at the beginning of the treatment, we suppose that at the beginning the larger weight of men causes larger instantaneous elastic deformations. However, lumbar segments of women must have poorer damping capacity, so creep needs shorter time to develop, since the final deformations of women quasi-overtake these of the men. Moreover, the effect of certain psychical differences between men and women cannot be excluded (capacity of self-accommodation and relaxation, temporary fear of water, etc.). The results of creep analysis based on the measured elongations, concerning the behaviour of men and women, are shortly reported in [13]–[16] and will be detailed in forthcoming papers.

16. It has been numerically verified that tensile deformability of human lumbar segments is inversely proportional to the *aging*: deformations decrease with increasing

age of patients in both the instantaneous elastic and the time-dependent creeping phases of the traction therapy. In the group with extra weights, patients having deformations immediately at suspension or at any time of the treatment were significantly younger than patients with undeformed segments. In the group with no extra weights, patients had more rigid and degenerated segments, thus, this observation was manifested as a whole, except for LV-SI where no age correlation could be registered. For the patients with obstinate rigidity, mean age was increasing in time: undeformed segments belonged to even older and older patients. These observations support the results of ACAROGLU et al. [1] who studied human lumbar annulus fibrosus to evaluate the effect of aging and degeneration on the tensile properties. IATRIDIS et al. [10] have found a statistically significant increase in the shear moduli of the nucleus pulposus with increasing age and extent of degeneration. They conclude that the nucleus pulposus undergoes a transition from fluid-like behaviour to more solid-like behaviour with aging and degeneration. TWOMEY and TAYLOR [27] have proved that the discs sink into the vertebrae with aging. Thus, with aging, obviously also for traction load, smaller deformations may occur. However, by analyzing the effect of aging on deformability, the fact has to be considered as well that aging is definitely negatively correlated with body height: with increasing age the body height decreases. Since shorter segment has smaller deformation, thus, besides the increasing rigidity due to the material aging, also the geometrical shortening causes the reduced deformability of segments. In the numerical model of segments, these complex effects yield a higher-order approximation for trend lines of age-sensitivity. By using linear approximation of each phase of treatment, we have found that the slope of trend lines increases in time, thus, the age-sensitivity seems to be increasing in time. Aging effects are reported shortly in [12], [14], [15], more details in forthcoming papers.

17. We have observed that not only the ratio of rigidity is more significant in *distal direction* but also the effect of aging. Similarly, ADAMS et al. [2] showed that age-related degenerative changes reduced the diameter of the functional nucleus and increased the width of the functional annulus; moreover, the effects of age and degeneration were greater at L4-L5 than at L2-L3.

18. In the group with extra loads, the tensile deformability of human lumbar segments increased with increasing *body height* both at the beginning and in the further phases of the traction therapy. This is partly due to the fact that a body height is proportional to the original length of segments, and a longer segment shows larger elongation. On the other hand, as mentioned above, a body height is negatively correlated with aging: during aging the body height decreases, thus taller patients are generally younger. These complex effects allowed us to assume a second-order approximation for the trend lines of functions of height sensitivity. Height dependence showed similar tendency in all time of the treatment. Height and aging effects need a common analysis by introducing a combined biomechanical parameter containing both of them.

19. There is no unambiguous correlation between the tensile deformability of lumbar segments and the *body weight*. For the group with extra loads, at the beginning of the treatment, deformations increased slightly with increasing body weight. Later on, and at the end of the treatment, this correlation could not be observed any more. What is more, observable differences between men and women appeared in the weight effects. This fact supports the assumption that the different weight and damping features of male and female patients are manifested themselves as creeping behaviour, yielding different deformation propagation process in time. Thus, the effect of body weight needs more analysis related to the creeping behaviour of segments. Preliminary results can be found in [14], [16], a detailed analysis will be reported later.

20. According to the measurements *just after the treatment*, in the patients leaving a basin, 80–90% of segment elongations seem to be disappeared, or rather the remaining deformations are smaller than the observation limit. This is due to the rapid re-compression again. However, during some weeks with daily repeated treatments, these small deformations can be large enough to allow the lumbar joint being treated to recover.

According to a remarkable prophecy of NACHEMSON [24], the conservative treatments should currently enjoy a renaissance in low back pain problems. In his opinion, for the majority of patients suffering from low back pain and showing no need for surgical treatment, help should come due to collaborative efforts of – among others – engineers, politicians, physicians and biomechanicians. The prophecy seems to be verified during the last two decades. Even the present supported study aimed to improve the conservative treatments of low back pain based on the biomechanical efforts of engineers and physicians.

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Refecences

[1] ACAROGLU E.R., IATRIDIS J.C., SETTON L.A., FOSTER R.J., MOW V.C., WEIDENBAUM M., *Degeneration and aging affect the tensile behaviour of human lumbar anulus fibrosus*, Spine, 1995, **20**(24), 2690–2701.

[2] ADAMS M.A., MCNALLY D.S., DOLAN P., *Stress distributions inside intervertebral discs. The effects of age and degeneration*, J. Bone Joint Surg. Br., 1996, **78**(6), 965–72.

[3] ANDERSSON G.B., SCHULTZ A.B., NACHEMSON A.L., *Intervertebral disc pressures during traction*, Scand. J. Rehabil. Med. Suppl., 1983, **9**, 88–91.

[4] BADER D.L., BOUTEN C., *Biomechanics of soft tissues*, [in:] *Clinical Biomechanics*, Editor: Z. Dvir, Churchill Livingstone, New York, London, 2000, 35–64.

[5] BENE É., *Das Gewichtbad*, Z. Phys. Med. Baln. Klim., 1988, **17**, 67–71.

[6] BENE É., KURUTZ M., *Application and biomechanics of traction bath* (in Hungarian), Orvosi Hetilap, 1993, **134**(21), 1123–1129.
[7] BOOS N., WALLIN A., AEBI M., BOESCH C., *A new magnetic resonance imaging analysis method for the measurement of disc height variations*, Spine, 1996, **21**, 563–570.
[8] BOTSFORD D.J., ESSES S.I., OGILVIE-HARRIS D.J., *In vivo diurnal variation in intervertebral disc volume and morphology*, Spine, 1994, **19**, 935–940.
[9] CHEN Y.G., LI F.B., HUANG C.D., *Biomechanics of traction for treatment of lumbar disc prolapse* (in Chinese), Chung Hua I Hsueh Tsa Chih,1994, **74**, 40–42.
[10] IATRIDIS J.C., SETTON L.A., WEIDENBAUM M., MOW V.C., *Alterations in the mechanical behavior of the human lumbar nucleus pulposus with degeneration and aging*, J. Orthop. Res., 1997, **15**(2), 318–22.
[11] KURUTZ M., LOVAS A., TORNYOS Á., MOLNÁR P., MONORI E., *Experimental biomechanical tensile model of human lumbar spine segments in vivo*, Proceedings of the Fifth World Congress on Computational Mechanics, (WCCM V), July 7–12, 2002, Vienna, Austria, Editors: H.A. Mang, F.G. Rammerstorfer, J. Eberhardsteiner, http://wccm.tuwien.ac.at, pp. 1–10.
[12] KURUTZ M., LOVAS A., MOLNÁR P. MONORI E., *The effect of aging in time-dependent tensile deformability of human lumbar spine segments in vivo*, Acta of Bioengineering and Biomechanics, 2002, **4**(Suppl. 1), 92–93.
[13] KURUTZ M., BENE É., LOVAS A., MONORI E., MOLNÁR P., *In vivo time-dependent tensile deformability of human lumbar spine segments, measured in traction bath therapy in terms of aging, sexes, weight and height*, Proceedings of IV World Congress of Biomechanics, CD, Calgary, Canada, August 4–9, 2002.
[14] KURUTZ M., *Age- and sex-dependent in vivo viscoelastic numerical tensile models of human lumbar spine segments*, Proceedings of IV World Congress of Biomechanics, CD, Calgary, Canada, August 4–9, 2002.
[15] KURUTZ M., BENE É., LOVAS A., MONORI E., MOLNÁR P., *Biomechanical parameter analysis of time-dependent tensile deformability of human lumbar spine segments in vivo*, Proceedings of the 2nd European Medical and Biological Engineering Conference, EMBEC'02, http://www.embec.org/, Vienna, Austria, December 4–8, 2002.
[16] KURUTZ M., *Parameter-dependent in vivo viscoelastic numerical tensile models of human lumbar spine segments*, Proceedings of the 2nd European Medical and Biological Engineering Conference, EMBEC'02, http://www.embec.org/, Vienna, Austria, December 4–8, 2002.
[17] LEBLANC A.D., EVANS H.J., SCHNEIDER V.S., WENDT R.E. 3rd, HEDRICK T.D., *Changes in intervertebral disc cross-sectional area with bed rest and space flight*, Spine, 1994, **19**, 812–817.
[18] LEDSOME J.R., LESSOWAY V., SUSAK L.E., GAGNON F.A., GAGNON R., WING P.C., *Diurnal changes in lumbar intervertebral distance, measured using ultrasound*, Spine, 1996, **21**, 1671–1675.
[19] MALKO J.A., HUTTON W.C., FAJMAN W.A., *An in vivo magnetic resonance imaging study of changes in the volume (and fluid content) of the lumbar intervertebral discs during a simulated diurnal load cycle*, Spine, 1999, **24**, 1015–1022.
[20] MOLL K., *Treatment of discushernia with so-called "weightbath" therapy* (in Hungarian), Orvosi Hetilap, 1953, **94**, 292–296.
[21] MOLL K., *Die Behandlung der Discushernien mit den sogenannten „Gewichtsbadern"*, Contempl. Rheum., 1956, **97**, 326–29.
[22] MOLL K., *On the new and simple treatment of discushernia* (in Hungarian), Orvosi Hetilap, 1957, **98**, 531–532.
[23] MOLL K., *The role of traction therapy in the rehabilitation of discopathy*, Rheum. Balneol. Allerg., 1963, **3**, 174–177.
[24] NACHEMSON A.L., *Advances in low-back pain*, Clin. Orthop., 1985, **200**, 266–278.
[25] RAMOS G., MARTIN W., *Effects of vertebral axial decompression on intradiscal pressure*, J. Neurosurg., 1994, **81**, 350–353.

[26] ROBERTS N., HOGG D., WHITEHOUSE G.H., DANGERFIELD P., *Quantitative analysis of diurnal variation in volume and water content of lumbar intervertebral discs*, Clin. Anat., 1998, **11**, 1–8.
[27] TWOMEY L., TAYLOR J., *Age changes in lumbar intervertebral discs*, Acta Orthop. Scand., 1985, **56**(6), 496–9.