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The effects of creep and recovery on the *in vitro* biomechanical characteristics of human multi-level thoracolumbar spinal segments

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A B S T R A C T

Background: Several physiological and pathological conditions in daily life cause sustained static bending or torsion loads on the spine resulting in creep of spinal segments. The objective of this study was to determine the effects of creep and recovery on the range of motion, neutral zone, and neutral zone stiffness of thoracolumbar multi-level spinal segments in flexion, extension, lateral bending and axial rotation.
Methods: Six human cadaveric spines (age at time of death 55–84 years) were sectioned in T1–T4, T5–T8, T9–T12, and L1–L4 segments and prepared for testing. Moments were applied of +4 to −4 N m in flexion-extension, lateral bending, and axial rotation. This was repeated after 30 min of creep loading at 2 N m in the tested direction and after 30 min of recovery. Displacement of individual motion segments was measured using a 3D optical movement registration system. The range of motion, neutral zone, and neutral zone stiffness of the middle motion segments were calculated from the moment-angular displacement data.
Findings: The range of motion increased significantly after creep in extension, lateral bending and axial rotation ($P<0.05$). The range of motion after flexion creep showed an increasing trend as well, and the neutral zone after flexion creep increased by on average 36% ($P<0.01$). The neutral zone stiffness was significantly lower after creep in axial rotation ($P<0.05$).
Interpretation: The overall flexibility of the spinal segments was in general larger after 30 min of creep loading. This higher flexibility of the spinal segments may be a risk factor for potential spinal instability or injury.

1. Introduction

During normal daily activities, the spine is subjected to a combination of compression, bending, and torsion that may be of a static or dynamic nature. Several physiological conditions in daily life cause sustained static bending or torsion loads on the spine. The clearest example is prolonged sitting of which Adams and Dolan (1991) stated that a bending moment of already 2–4 N m occurs on the lumbar spine during a sustained flexion of only 20°. For a better understanding of the biomechanics of the spine and of the consequences of these common sustained loads it is important to know how these loads affect the instantaneous biomechanical behaviour of the spine.

During sustained loading of the spine, creep occurs. Twomey and Taylor (1982) defined creep as the progressive deformation of a structure under constant load when the materials are stressed well below their fracture thresholds. They described creep behaviour of the spine in flexion and showed the deformation of a multi-level spinal segment during a period of static loading. Deformation during a recovery period was much more rapid, but full recovery was not achieved during the test time.

Several authors (Adams and Dolan, 1991; Adams and Dolan, 1996; Little and Khalsa, 2005; McGill and Brown, 1992) have provided additional data describing the creep behaviour of the spine in flexion or extension. However, it is still largely unknown how the instantaneous biomechanical characteristics of spinal segments, such as the range of motion (RoM), neutral zone (NZ), and neutral zone stiffness (NZStiff), are affected by sustained static loading.

Adams and Dolan (1996) and Little and Khalsa (2005) were the only authors who measured instantaneous kinematics of a spinal motion segment after creep. They found a significant increase in range of motion after flexion creep, as well as after a recovery period.

In line with this finding, it is generally believed that lumbar flexion creep results in a laxity of the intervertebral joint, the ligaments, capsules, and discs (Adams and Dolan, 1991; Little and Khalsa, 2005; Goel et al., 1988), which would influence the motion behaviour of the

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lumbar spine. This is plausible as well for other creep directions, and therefore bending or rotational creep is expected to result in a larger range of motion, a larger neutral zone, and a smaller neutral zone stiffness.

Existing literature only describes creep behaviour in flexion and extension directions of the lumbar region of the spine. Due to differences in geometry of the intervertebral joints, ligaments, and discs between spine regions, it can be questioned whether the effects of creep in the thoracic or lumbar regions of the spine are different. No information on *in-vitro* creep in lateral bending or axial rotation, and of thoracic regions of the spine could be found. Furthermore, most studies only tested single motion segments, whereas a multi-level segment appears to be more representative for the normal physiological situation (Adams, 1995; Dickey and Kerr, 2003; Goel et al., 2006).

The goal of this study, therefore, was to determine the effects of creep and recovery on the range of motion, neutral zone, and neutral zone stiffness of thoracolumbar multi-level spinal segments in flexion, extension, lateral bending, and axial rotation.

2. Methods

2.1. Specimen and specimen preparation

Six fresh frozen spines were used in this study. The specimens were harvested from 6 male cadavers obtained from the Department of Anatomy of the University Medical Center Groningen, The Netherlands. Mean age of the subjects at the time of death was 72 years (range 55–84 years).

The spines were dissected from T1–L4 and all musculature was carefully removed leaving the ligaments intact. At both lateral sides of the spine approximately 3 cm of ribs was preserved, including the costotransverse and costovertebral articulations. CT scans of the spines were made and showed mildly degenerated but otherwise normal specimens without radiological evidence of spinal pathology.

Each spine was sectioned into multi-level segments containing 4 vertebrae and 3 intervertebral discs: T1–T4, T5–T8, T9–T12, and L1–L4. These segments were stored at −40 °C prior to testing. Eighteen hours before testing, the segments were thawed and three screws were driven into the endplate of the upper and lower end-vertebrae to improve the fixation. The upper and lower end-vertebrae were embedded in cups fitting in the test setup, using a low melting temperature alloy (Cerrolow-147; 48% bismuth, 25.6% lead, 12% tin, 9.6% cadmium, and 4% indium). All articulating parts were kept free.

A marker containing three LEDs was rigidly fixed to the anterior surface of each vertebral body (Fig. 1). To minimize dehydration, saline-soaked gauze was wrapped around the spinal segments (not seen in Fig. 1) and the segments were sprayed with saline solution during preparation and testing. All tests were performed at room temperature.

2.2. Experimental test setup

A custom-made four-point bending device was used in this study in which flexion (FL), extension (EX), right and left lateral bending (LB), and right and left axial rotation (AR) could be applied (Fig. 1a and b). The bending device was driven by a Zwick mechanical material testing system (model TC-FR2.5TN, Zwick Roell, Ulm Germany, not seen in Fig. 1a and b). For flexion–extension and lateral bending, the load was applied to the two most lateral bending points of the test setup. The black beam in Fig. 1a and b represents the original beam which was attached to the Zwick material testing system. The rotated arrows indicate the bending points of the test setup. By using a four-point bending device, a continuous moment could be applied over the complete spinal segment (Fig. 1b). Due to the counterweights at the endings of the test-setup, a downward moment was existent when zero-load was applied. Therefore, positive and negative moments

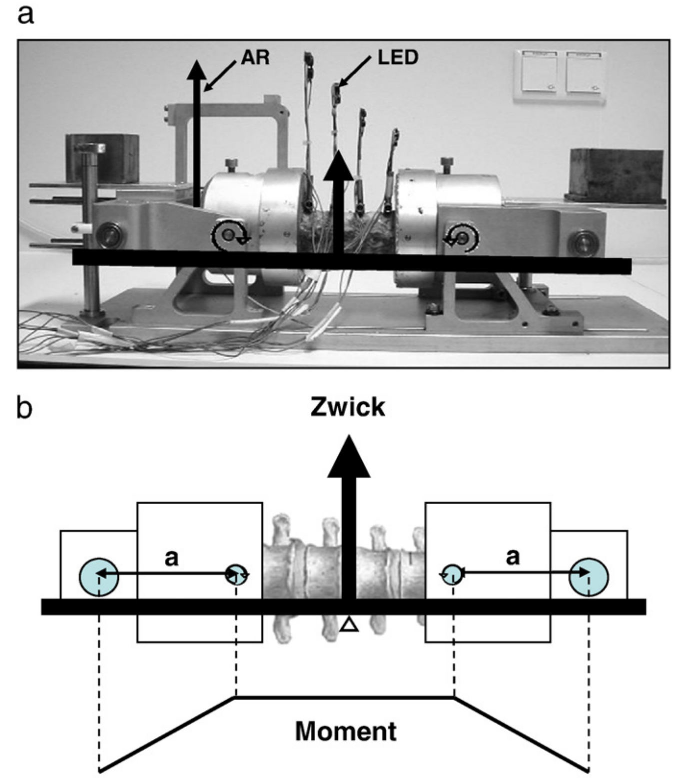

Fig. 1. a. Test setup for the application of moments to the multi-level spinal segments. A mechanical testing system applied loads at the points indicated by arrows. The specimen was rotated 90° for lateral bending. For axial rotation the left cup was rotated with a small steel cable driven by the same mechanical testing system (AR). The markers with the LEDs (LED) were rigidly fixed to the vertebrae. An Optotrak® camera registered the movement of the LEDs. b. Schematic view of the test setup for the application of moments to the multi-level spinal segments. Due to the four-point bending device, a continuous moment was present in between the two inner bending points. a = moment-arm.

could be applied in one continuous motion. The specimen was rotated 90° about its cranio-caudal axis in the test setup for testing lateral bending. Since the Zwick applied a linear force, the slight change of moment-arm during the rotation of the specimens was accounted for in the data-analysis. Axial rotation was applied via a steel cable attached to the left cup in the testing device, also driven by the Zwick testing system (AR in Fig. 1). A different counterweight was attached to the left cup to be able to test left and right axial rotations in one continuous motion, again with a continuous moment over the spinal segment (not seen in Fig. 1a and b).

The left cup in the test setup could always move without restraint in the longitudinal direction and axial rotation (the latter only when applying flexion-extension and lateral bending).

Before testing, an axial preload of 250 N was applied to the specimen for 1 h to minimize effects of superhydrated intervertebral discs. Preload was then removed and a rotation of 0.5°/s was applied to the specimen in the tested direction. When a force corresponding to +4 N m was measured by the load cell of the Zwick, the displacement direction was reversed until a force corresponding to −4 N m was reached. To precondition the specimens to their physiological range, specimens were tested for 3 continuous cycles (Wilke et al., 1998a,b). Although negligible differences were found between the 3 loading cycles, the recommendations of Wilke et al. (1998a,b) were followed and the data of the third cycle were analysed. After the three loading cycles, a continuous bending moment of 2 N m (load-controlled) was applied in one out of four creep directions (see below) for 30 min. Then the 3 loading cycles between +4 and −4 N m were repeated, followed by a recovery period of 30 min during which no load was applied. After the recovery period, again 3 loading cycles between +4 and −4 N m were applied. Subsequently, this procedure was repeated three more times for the other creep directions. Loads were applied in the following directions: (1) cyclical flexion–extension with creep loading in flexion; (2) cyclical flexion–extension with creep loading in extension, (3) cyclical left and right lateral bending with creep loading in one of the lateral bending directions, and (4) cyclical left and right axial rotations with creep loading in one of the axial rotation directions.

To neutralize effects of testing order, directions were applied in a different order for each individual segment.

Motions of the LEDs were recorded by an optoelectronic 3D movement registration system with one array of 3 cameras (Optotrak 3020, Northern Digital Inc, Waterloo ON). The sample rate was 50 samples/s. Before testing, the axes of the Optotrak system were aligned with the anatomic axes of the spinal segment.

2.3. Data analysis

Moment-angular displacement curves were generated for each cyclic test and biomechanical parameters were extracted from the third loading cycle. A computer program written in Matlab (Mathworks, Natick MA, USA) was used to track the axes system of each individual vertebra during the loading-cycle. Motion of intervertebral joints was determined by calculating the rotation of one vertebra relative to its neighbour in local axes. Data of the middle intervertebral joint, calculated from the middle two vertebrae, were used for analysis because this motion segment could move without restraint and was therefore more representative for the human physiological situation in comparison to the outer vertebral joints which were partially embedded in the test setup. Rotation in the local main (intended) direction, i.e. either flexion–extension, or lateral bending, or axial rotation, further denoted as angular displacement, was plotted as a function of load moment in order to obtain moment-angular displacement curves.

The range of motion (RoM), neutral zone (NZ), and neutral zone stiffness (NZstiff) of each motion segment were calculated from the moment-angular displacement data (Fig. 2). Range of motion was defined as the range of angular displacement between −4 N m and +4 N m. The neutral zone was calculated according to a recently developed method by Smit et al. (2009) in which the NZ is determined as the zone in between the points of the largest changes in flexibility in the moment-angular displacement curve. Stiffness in the NZ was calculated as the inverse of the slope in the complete range of the NZ, since this part of the curve showed linear characteristics (Fig. 2).

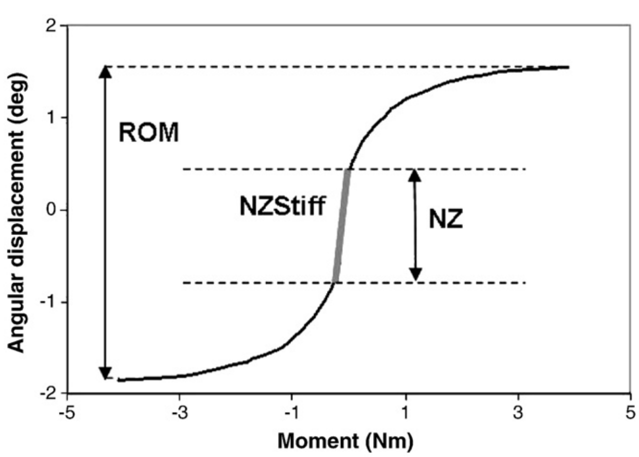

Fig. 2. Example of a moment-angular displacement curve. The RoM and NZ are shown. NZStiff is calculated as the inverse slope of the curve in the NZ.

To account for the intrinsic variability between segments of different spines, the parameters (RoM, NZ, and NZStiff) were normalized to the first measurement (pre-creep third cycle) in the tested direction. In separate ANOVAs for creep and recovery, those normalized values were tested for deviation from 1, i.e. from the pre-creep measurement, using the *P*-values of the model intercept of a univariate general linear model (SPSS Inc, Chicago USA). The dependent variables were the normalized RoM, NZ, or NZStiff. Fixed factor was the spinal region (T1–T4, T5–T8, T9–T12, and L1–L4) and the random factor was the individual spine number. If the ANOVA showed an effect of the spinal region, the effects of creep and recovery were tested *post hoc* with t-tests against the value 1, i.e. the pre-creep measurement, for each region separately. Differences were considered significant when $P<0.05$.

3. Results

Absolute values of RoM, NZ, and NZStiff before creep have been published elsewhere (Busscher et al., 2009). The results described here will focus on the effects of creep and recovery on these absolute values. The results for each normalized parameter after creep and recovery in flexion (FL), extension (EX), lateral bending (LB), and axial rotation (AR) are shown in Figs. 3–5.

The RoM was significantly larger after 30 min of creep in extension, lateral bending, and axial rotation ($P<0.05$). The effects were relatively small, as RoM increases were, at all vertebral levels, on average 10% or less (Fig. 3). Fig. 3 suggests a larger increase of RoM after flexion creep, but this increase did, probably due to large standard deviations, not reach significance. RoM was still significantly larger after recovery in lateral bending $P<0.05$).

The NZ was significantly larger after creep in flexion, with an increase of on average 36.1% (SD 47.2), and did not return to the baseline values during the recovery period (Fig. 4).

On average, the NZStiff was significantly smaller after creep in axial rotation only, particularly in the more cranial thoracic and lumbar regions. Averaged over regions, NZStiff did return to baseline values after recovery (Fig. 5).

There was a significant effect of spinal region for RoM in axial rotation after rotation-creep and NZStiff in flexion–extension after extension-creep ($P<0.05$). Testing the individual regions in these cases showed a significant effect of creep on the RoM in axial rotation for the lumbar region whereas the RoM in the thoracic regions was not significantly affected by creep (Fig. 3) The NZStiff for flexion–extension was significantly smaller after extension-creep in the thoracic regions, but not in the lumbar region (Fig. 5).

4. Discussion

In this study, the effects of creep and recovery on the biomechanical characteristics of thoracolumbar spinal segments were determined. This is the first study that included thoracic spinal segments as well as lateral bending and axial rotation. Furthermore, this study investigated the consequences of common, physiological creep loading, which occurs during many daily activities.

A significant increase was found for the RoM after creep in extension, lateral bending, and axial rotation. The NZ was significantly larger after creep in flexion, and NZStiff was significantly smaller after creep in axial rotation. It was anticipated that different effects of creep would be present in various regions of the spine due to differences in geometry of the vertebrae, joints, and intervertebral discs. However, this study showed that for most motion directions, the effects of creep and recovery with the present loading magnitudes were comparable between spinal regions. Only the effect on the RoM in axial rotation was different between thoracic and lumbar spinal regions. Possibly, this was an effect of the normalisation of the values. The absolute RoM in the lumbar region

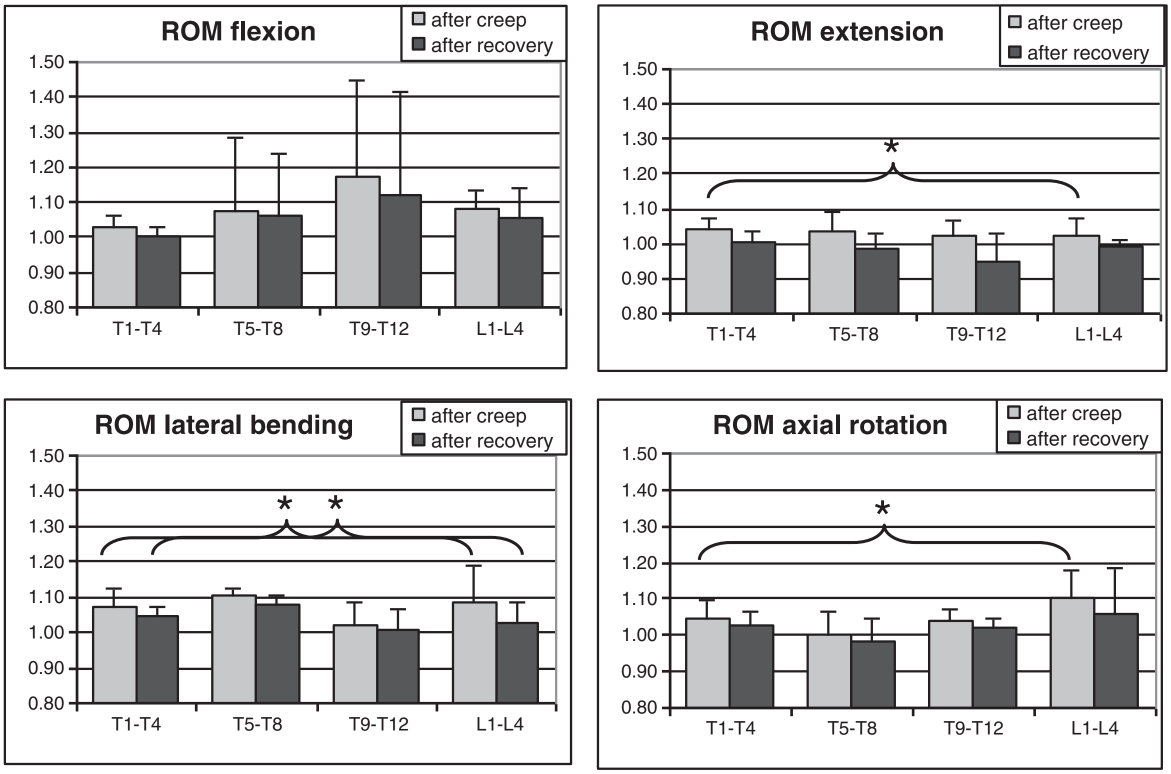

Fig. 3. Average normalized RoM data of the third loading cycle after creep and after recovery. Shown are flexion–extension RoM after flexion creep, flexion-extension RoM after extension creep, left–right lateral bending RoM after one-sided lateral bending creep and left–right axial rotation RoM after one-sided axial rotation creep. The data are normalized to the third loading cycle before 30 min static loading and recovery (not shown in this figure). *Overall significant difference to the measurement before creep or recovery.

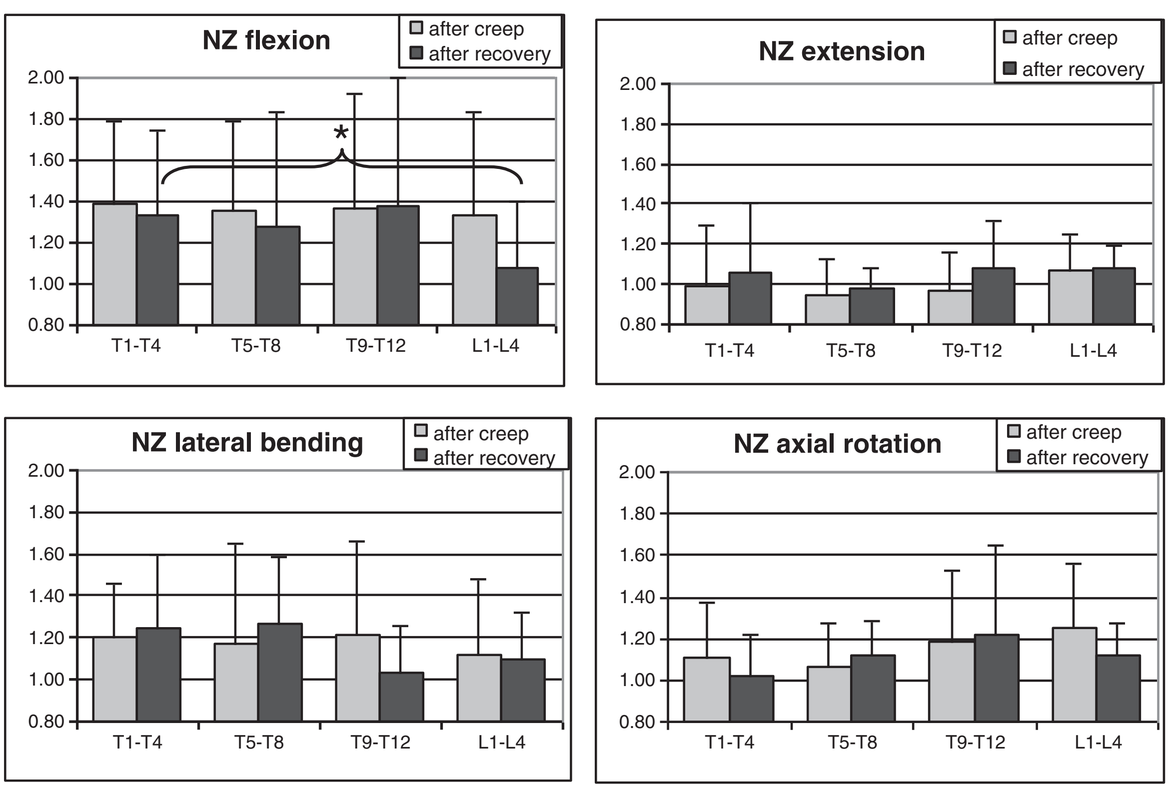

Fig. 4. Average normalized NZ data of the third loading cycle after creep and after recovery. Shown are flexion–extension NZ after flexion creep, flexion–extension NZ after extension creep, left–right lateral bending NZ after one-sided lateral bending creep and left–right axial rotation NZ after one-sided axial rotation creep. The data are normalized to the third loading cycle before 30 min of static loading and recovery (not shown in this figure). *Overall significant difference to the measurement before creep or recovery.

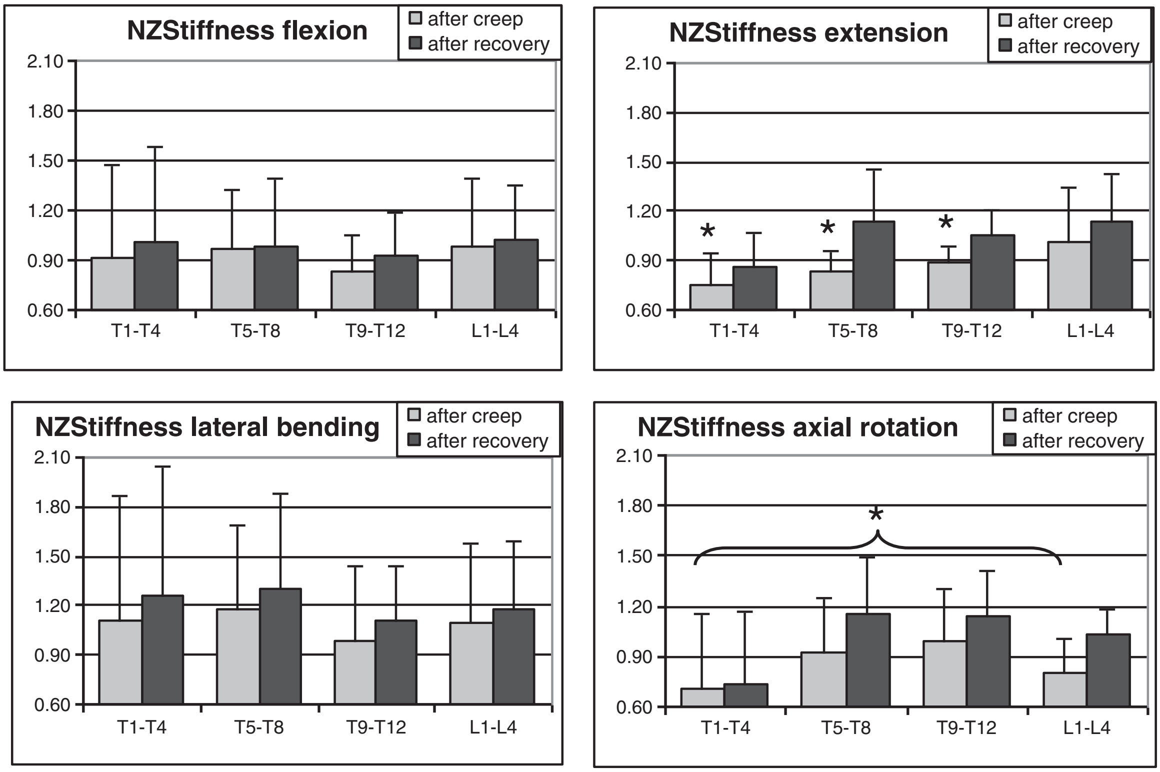

Fig. 5. Average normalized NZStiff data of the third loading cycle after creep and after recovery. Shown are flexion–extension NZStiff after flexion creep, flexion-extension NZStiff after extension creep, left–right lateral bending NZStiff after one-sided lateral bending creep and left–right axial rotation NZStiff after one-sided axial rotation creep. The data are normalized to the third loading cycle before 30 min static loading and recovery (not shown in this figure). *Overall significant difference to the measurement before creep or recovery.

was only 2.2° (Busscher et al., 2009), whereas the absolute RoM in the high thoracic region was 7.2°. A small change in the RoM in the lumbar region would therefore results in a larger relative effect of the creep. However, the exact influence of differences in geometry of the vertebrae, facet joints, and intervertebral discs should be investigated further.

A larger RoM indicates that the spinal segments were more flexible and potentially more likely to become unstable, even after a sustained loading at only 2 N m. During daily life, sustained thoracic and lumbar bending and torsion of this magnitude occur in many common activities like construction work, gardening, and household work. Sustained lumbar flexion is the most commonly occurring static load in daily life during forward bending and particularly during prolonged sitting. After sustained lumbar flexion the neutral zone increased by up to 30–40% for all spinal regions and 30 min of recovery did not appear to be sufficient to return to the baseline situation. This increase in magnitude of the NZ indicates a larger zone of potential instability and therefore more necessity for the muscles to compensate for this (Cholewicki et al., 1997). Panjabi (1992) and Oxland and Panjabi (1992) also stated that a larger magnitude of the neutral zone is a sensitive indicator for spinal instability and injury. Solomonow et al. (1999) stated that creep induces laxity in the ligaments, joint capsules, and intervertebral discs, and that, more important, this laxity desensitizes the mechanoreceptors within and results in a significant decrease or complete elimination of reflexive muscular stabilizing forces in the multifidus muscles. Furthermore, they showed that viscoelastic recovery *in-vivo* is slow (Gedalia et al., 1999), as seen as well in the present study. It is obvious that in the present *in-vitro* study muscular stabilizing forces are not present and therefore the results should be carefully extrapolated and interpreted for the human *in-vivo* situation. However, our results do suggest that prolonged loads, even if they are small, could potentially reduce spine stiffness *in-vivo*. This might increase the need for muscular compensation and increase the risk of injury.

A non-significant increase was observed in the RoM after flexion creep, in line with effects in the other three directions. The same held for the NZ after creep in lateral bending and axial rotation, which also showed a non-significant increase. Although a repeated measures design was used, the non-significance could be the result of a large variability of the parameters in the different spines, as overall the results indicated a decrease in stiffness after creep loading. This problem of a large variability over spines has previously been reported in *in-vitro* tests with human specimens (Twomey and Taylor, 1982; Wilke et al., 1998a,b; Smit, 2002). As in other studies, a limitation of this study was the small sample size. However, due to limited availability of human cadaver spines, no more specimens could be used.

Results of the present study should be interpreted taking the study protocol into account. Different studies use highly different protocols and recommendations for biomechanical testing of spinal segments are variable. In present study the recommendations of Wilke et al. were followed for length and preparation of the segments, test environment, and loading conditions. Wilke et al. (1998a,b) and Goel et al. (2006) recommended instantaneous loading between 3 and 7.5 Nm for the complete spine. A pilot study in our test setup showed damage or breakage of the higher thoracic segments when loaded to more than 5 N m in the cyclic tests. 4 N m appeared to be an appropriate amount to prevent breakage of the high-thoracic specimens and to still cause sufficient movement of the lumbar specimens. Furthermore, it was believed that 4 N m, as well as 2 N m sustained loading, was representative for common daily activities (Adams and Dolan, 1991).

Information regarding the amount of load during the creep phase is scarce. Adams (1995) recommended 10 N m for the lumbar spine but information for the thoracic spine could not be found. As we wanted to test all directions in each specimen of the four spinal regions, we anticipated that a 10 N m load would have created interpretation problems due to damage and subsequent order effects.

A load of 10 Nm appeared to be too high, particularly for the thoracic specimens. Furthermore, this study was aimed to test the effects of commonly occurring sustained loading which occurs, for example in prolonged sitting. Adams and Dolan (1991) showed that 20° of flexion already resulted in a moment of 2 N m, and this magnitude was therefore chosen for the creep loading phase in this study. Note that when flexion loads were combined with compression, much higher moments can be applied before damage occurs. Adams and Dolan (1991) tested their lumbar segments up to the elastic limit and found an average bending moment of 51.7 N m, but they tested under simultaneous compressive load.

Little and Khalsa (2005) tested the RoM in a lumbar segment after creep and recovery. A significant increase of 19% in the RoM was found after 20 min of flexion creep at 10 N m, and a residual creep of 11% remained after 20 min of recovery. Furthermore, Adams and Dolan (1996) showed that sustained flexion can reduce the lumbar spine's resistance to bending by 42% after only 5 min of creep loading between 25 and 35 N m. In the present study, the RoM increase after flexion creep did not reach significance but we creep loaded the segments at only 2 Nm.

Results of axial rotation and lateral bending presented in this study are difficult to compare with existing literature since no literature was found concerning the changes in these directions after creep and recovery. Furthermore, existing literature only reported data concerning creep of lumbar spinal segments.

Most studies describing flexion creep behaviour applied a constant load for only 20 min (Kaigle et al., 1992; Little and Khalsa, 2005; McGill and Brown, 1992). Twomey and Taylor (1982) however, showed there is still creep displacement after 20 min and a plateau phase was reached only after 25 min in the older specimens. We therefore chose to apply a moment for 30 min, and moment-angular displacement curves of the creep phase showed a plateau indeed. The recovery phase did not show a plateau in all specimens. Gedalia et al. (1999) and van der Veen et al. (2007) showed that the recovery after creep loading is very slow and only partial mechanical recovery takes place in the first min of the recovery period. Therefore, no full recovery as expected beforehand, except for some mechanical recovery in the spinal segments, as shown as well by Solomonow et al. (1999).

A limitation of the present study was that testing was not performed under compressive axial load. In the pilot work for our study, testing of flexion–extension and lateral bending under compressive load resulted in buckling and instability of the multi-level spinal segment. Some studies used a follower load (Patwardhan et al., 2003; Rohlmann et al., 2001), but as substantial motions occurred during testing, a follower load would have induced unintended additional moment loads on the segments of unknown direction and magnitude, as the axis of rotation of individual intervertebral joints is not precisely known and is not constant (Adams, 1995; Cripton et al., 2000). Therefore, we chose to only apply a compressive preload before testing to minimize the effect of superhydrated intervertebral discs.

Goel et al. (2006) recommended an axial load between 100 N and 400 N for the cervical to the lumbar spine and Zhao et al. (2005) used an axial load between 200 and 500 for the lumbar spine. Based on these values, we selected a 250 N preload for the whole spine. Furthermore, the pilot work showed that most of the effect of the preload was achieved in 1 h and therefore one hour preload was chosen.

Static compressive and bending loads each have a different influence on the intervertebral disc, and therefore on the motion behaviour of spinal segments (Stanley et al., 2004; Tawackoli et al., 2004; Wilke et al., 1998a,b). Zhao et al. (2005) showed the influence of compressive load on the RoM and NZ of lumbar spinal segments in flexion, extension, and lateral bending. After a compressive load of 1500 N for 2 h, the RoM increased by 30.6% in flexion, 6.3% in extension, and 39.6% in lateral bending. The NZ increased by 42.9%, 15.4%, and 71.4% respectively. This was probably mainly due to fluid loss from the nucleus pulposus. Van der Veen et al. (2007) showed that recovery of the intervertebral disc after compressive load lasted more than 12 h. In contrast to compression creep, bending creep most likely results in little fluid loss and substantial viscoelastic strain of the fibres of the annulus fibrosis. As the time constant for viscoelastic creep is lower than for fluid loss, the influence of creep as well as recovery on the instantaneous motion behaviour of spinal segments likely differs as well. Thus, besides applying multiple versus single tests, load magnitude and creep type may in part explain the disparity between the results of Zhao et al. (2005) and the present study.

Another limitation of the present study, as with all *in vitro* studies, was the non-physiological test-environment. The test environment and hydration status of the spinal segment influence the biomechanical characteristics during testing (Pflaster et al., 1997; Race et al., 2000). It is generally believed that wrapping the segments in moist gauze, and spraying them with saline during testing are sufficient to assure its moist condition, and to prevent large effects of overhydration, which can occur during testing in a saline bath (Wilke et al., 1998a,b).

A final limitation is that, for reasons of availability of cadaveric spines, we only tested spines of elderly subjects. Several studies (Twomey and Taylor, 1982; Kaigle et al., 1992; Oliver and Twomey, 1995) reported an effect of age on the amount of creep. These studies showed that more creep occurred in older specimens, and that it took longer to reach the plateau phase. Therefore, care should be taken in generalizing the present results to spines in younger subjects.

5. Conclusions

In conclusion, this study showed the influence of 30 min of creep loading and 30 min of recovery on the instantaneous behaviour in flexion–extension, lateral bending, and axial rotation of four thoracolumbar regions of the spine. An increase in RoM was found in all directions after creep loading, and an average increase of 36% was found of the neutral zone after flexion creep.

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