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Biomechanical contribution of spinal structures to stability of the lumbar spine—novel biomechanical insights

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Abstract

BACKGROUND CONTEXT: The contribution of anatomical structures to the stability of the spine is of great relevance for diagnostic, prognostic and therapeutic evaluation of spinal pathologies. Although a plethora of literature is available, the contribution of anatomical structures is still not well understood.

PURPOSE: We aimed to quantify the biomechanical relevance of each of the passive spinal structure through deliberate biomechanical test series using a stepwise reduction approach on cadavers.

STUDY DESIGN: Biomechanical cadaveric study.

METHODS: Fifty lumbar spinal segments originating from 22 human lumbar cadavers were biomechanically tested in a displacement-controlled stepwise reduction study: the intertransverse ligaments, the supraspinous and interspinous ligaments, the facet joint capsules (FJC), the facet joints (FJ), the ligamentum flavum (LF), the posterior longitudinal ligament (PLL), and the anterior longitudinal ligament were subsequently reduced. In the intact state and after each transection step, the segments were physiologically loaded in flexion, extension, axial rotation (AR), lateral bending (LB) and with anterior (AS), posterior (PS) and lateral shear (LS). Thirty-two specimens with only minor degeneration, representing a reasonably healthy subpopulation, were selected for the here presented evaluation. Quantitative values for load and spinal level dependent contribution patterns for the anatomical structures were derived.

RESULTS: Small variability between of the contribution patterns are observed. The intervertebral disc (IVD) is exposed to about 67% of the applied load in LB and during shear loading, but less by load in flexion, extension and AR (less than 35%). The FJ&FJC are the main stabilizers in AR with 49%, but provide only 10% of the stability in extension. Beside the IVD, the LF and the PLL contribute mainly in flexion (22% and 16%, respectively), while the ALL plays a major role during extension (40%) and also contributes during LB (15%). The contribution of the intertransverse ligaments and the supraspinous and interspinous ligaments are very small in all loading directions (<2% and <6%, respectively).

CONCLUSION: The IVD takes the main load in LB and absorbs shear loading, while the FJ&FJC stabilize AR. The ALL resists extension while LF and PLL stabilize flexion. With the small variability of contribution patterns, suggesting distinct adaptation of the structures to one another, the biomechanical characteristics of one structure have to be put in context of the whole spinal segment.
Introduction

The spine is a complex composition of various active and passive structures that allow for an erect posture, enable sophisticated movements and withstand loading during daily activities. While the active, neuromuscular components control movement, the passive structures of the spine provide stability, limit and define range of motion and protect the neural structures in the spinal canal. Apart from the vertebrae, the main anatomical structures are the intervertebral disc (IVD) connecting the vertebral corpuses, the anterior longitudinal ligament (ALL) covering the ventral section of the vertebrae, the posterior longitudinal ligament (PLL) which is attached to the dorsal cortex of the vertebra, the ligamentum flavum (LF) connecting the laminae of the spinal column, the intertransverse ligaments (ITL) linking the transversal processes and the interspinous and supraspinous ligaments (ISL&SSL) that interconnect the dorsal lying spinous processes. The only bony structure that provides passive stability are the facet joints (FJ) composed of cartilaginous joint surfaces surrounded by the facet joint capsules (FJC). Due to their anatomical location and specific compositions, each passive structure contributes to the stability during certain movements and under certain loading conditions. Due to the resulting mechanical stresses, these structures are prone to degeneration, which can alter their effectiveness and lead to pain or malfunction. It is therefore essential to understand the particular function of each of these spinal structures. Furthermore, comprehensive knowledge is needed to understand clinical presentations of specific medical conditions, to comprehend the pathomechanisms in degenerative and traumatic diseases, to optimize surgical interventions and to accurately model biomechanical finite element and multibody dynamics simulations.

In this context, several “stepwise transection studies” were performed in the past. Bending moments were applied to spinal segments in vitro, structures were sequentially removed and the resulting increase in range of motion (ROM) was measured [1–8]. In the most recent and extensive study the ISL, LF, FJ, dorsal arch, PLL, ALL, and nucleus were resected in 8 spinal segments with biomechanical loading in flexion, extension, lateral bending (LB) and axial rotation (AR) [3]. This study gave some profound insight in the importance of the passive stabilizing structures. However, certain drawbacks are associated with such a load-controlled experiment design. Since progressively fewer structures are exposed to the same loads, overloading of the remaining structures resulting in plastic deformation or specimen failure cannot be excluded. To prevent overloading, a displacement-controlled biomechanical testing protocol should be used [9]. With this approach, the influence of the spinal structures can be determined by measuring the reduction in load required for the same motion. However, specific demands to the setup have to be met: Since loading is displacement-controlled, minimal errors in the starting position could lead to a large error in the measurements, especially since load-deflection curves of spinal segments are usually progressive. Similarly, uncontrolled slippage of the specimens within the clamping system could interfere with findings and places great demands on the clamping technique. Furthermore, viscoelastic effects have to be controlled to exclude progressive changes in stiffness of the spinal segments [10]. Likewise, dehydration over the duration of the experiments has to be prevented [10]. The goal of this study was to develop a setup that is able to bring all of these parameters under suitably tight control to allow a displacement-controlled transection study to investigate the load apportionment among the passive structures of the lumbar spine.

Material and methods

Dissection, preparation, and storage

This study was approved by the local ethical authorities (BASEC Nr. 2017-00874). Fifty spinal segments originating from 22 fresh frozen cadavers (Table 1; Science Care, Phoenix, AZ, USA) were tested. After thawing, CT, MRI (3T and 7T), and planar X-rays scans were performed to exclude bony defects and other spinal pathologies. Only specimens with little to moderate degeneration were included. The specimens were carefully dissected without harming bony processes, paraspinal ligaments or the IVDs. After preparation, the segments were mounted with individualized 3D-printed-clamps specifically developed and validated for this study [11]. Compared with the standard potting procedure, this method prevents specimen heating, allows for precise specimen orientation, permits access to the dorsal structures and reduces slippage in the fixation.

Biomechanical test setup

Biomechanical testing was performed on a biaxial (linear & torsion) static testing machine (Zwick/Roell Allroundline 10kN and testXpert III Software, ZwickRoell GmbH & Co. KG, Germany). The system is based on a traverse to generate vertical compression and tension and a
torsion motor to generate torque in the horizontal plane (Fig. 1A). This machine was complemented with a testing setup consisting of an x-y-table and holding arms that allow for specimen fixation in a vertical orientation for axial compression-decompression (AC) and AR (Fig. 1D), in a horizontal orientation for LB, and AS (Fig. 1C) as well as flexion-extension (FE) and lateral shear (LS) (Fig. 1B). A customized mounting fixture for the clamped specimens, consisting of high precision fitting rings, pins and a mechanism to compress the connection with a defined load before

Table 1
Specimen table with all 32 includes samples

| Specimen | Level | Sex  | Age  | Cause of death          | Height [cm] | Weight [kg] | BMI  | Pfirrmann |
|----------|-------|------|------|--------------------------|-------------|-------------|------|-----------|
| S180106  | L1L2  | female | 71   | Cardiac Arrest            | 165.1       | 88.5        | 32.5 | 3         |
| S180088  | L3L4  | male  | 69   | Cardiac Arrest            | 187.96      | 176.9       | 50   | 4         |
| S173430  | L1L2  | male  | 65   | Acute Respiratory Failure | 170.18      | 121.6       | 42   | 4         |
| S173430  | L3L4  | male  | 65   | Acute Respiratory Failure | 170.18      | 121.6       | 42   | 3         |
| S180093  | L1L2  | male  | 69   | Pancreatic Cancer         | 182.88      | 98.9        | 29.6 | 4         |
| S180093  | L3L4  | male  | 69   | Pancreatic Cancer         | 182.88      | 98.9        | 29.6 | 3         |
| S180968  | L2L3  | male  | 82   | Cardiopulmonary Arrest    | 177.8       | 81.6        | 25.8 | 4         |
| S180968  | L4L5  | male  | 82   | Cardiopulmonary Arrest    | 177.8       | 81.6        | 25.8 | 3         |
| L181252  | T12L1 | male  | 66   | Alcoholic Liver Cirrhosis | 175.26      | 64.4        | 21   | 3         |
| L181252  | L2L3  | male  | 66   | Alcoholic Liver Cirrhosis | 175.26      | 64.4        | 21   | 4         |
| L181252  | L4L5  | male  | 66   | Alcoholic Liver Cirrhosis | 175.26      | 64.4        | 21   | 4         |
| S180718  | T12L1 | female | 73   | Coronary Artery Disease   | 170.18      | 113.4       | 39.2 | 3         |
| S181260  | L1L2  | male  | 66   | Cardiac Arrest            | 182.88      | 95.3        | 28.5 | 2         |
| S181260  | L3L4  | male  | 66   | Cardiac Arrest            | 182.88      | 95.3        | 28.5 | 3         |
| S181190  | L4L5  | male  | 83   | Congestive Heart Failure  | 187.96      | 87.5        | 24.8 | 4         |
| S181915  | T12L1 | male  | 66   | Cardiovascular Disease    | 177.8       | 86.2        | 27.3 | 3         |
| S181915  | L4L5  | male  | 66   | Cardiovascular Disease    | 177.8       | 86.2        | 27.3 | 3         |
| S181997  | L3L4  | male  | 82   | Coronary Artery Disease   | 185.42      | 90.7        | 26.4 | 4         |
| L180769  | L1L2  | male  | 86   | Pancreatic Cancer         | 175.26      | 49.9        | 16.2 | 3         |
| L180769  | L3L4  | male  | 86   | Pancreatic Cancer         | 175.26      | 49.9        | 16.2 | 4         |
| S182452  | L3L4  | male  | 62   | Acute Myocardial Infarction| 172.72      | 79.4        | 26.6 | 4         |
| S182571  | L1L2  | female | 84   | Respiratory Failure       | 165.1       | 66.7        | 24.5 | 3         |
| S182571  | L3L4  | female | 84   | Respiratory Failure       | 165.1       | 66.7        | 24.5 | 4         |
| S182575  | L1L2  | male  | 63   | Respiratory Arrest; COPD  | 172.72      | 108.9       | 36.5 | 2         |
| S182575  | L3L4  | male  | 63   | Respiratory Arrest; COPD  | 172.72      | 108.9       | 36.5 | 2         |
| S182576  | T12L1 | male  | 75   | Cardiac arrest            | 177.8       | 89.8        | 28.4 | 3         |
| S182576  | L2L3  | male  | 75   | Cardiac arrest            | 177.8       | 89.8        | 28.4 | 3         |
| S182664  | T12L1 | male  | 75   | Respiratory Arrest; COPD  | 185.42      | 97.5        | 28.4 | 3         |
| S182664  | L4L5  | male  | 75   | Respiratory Arrest; COPD  | 185.42      | 97.5        | 28.4 | 3         |
| S182898  | L4L5  | female | 51   | End Stage Liver Disease   | 172.72      | 100.7       | 33.8 | 2         |
| S190353  | T12L1 | male  | 48   | Malignant Neoplasm        | 177.8       | 88.9        | 28.1 | 3         |
| S190353  | L4L5  | male  | 48   | Malignant Neoplasm        | 177.8       | 88.9        | 28.1 | 2         |

70.2  176.8  90.7  29  3.2

Fig. 1. The setup for biomechanical testing (A) used to test spinal segments in flexion-extension and lateral shear (B), lateral bending and anteroposterior shear (C) and axial rotation and axial compression-decompression (D).
tightening was developed. This installation allowed for specimen fixation with extremely high reproducibility (variability <0.005°). Loading was applied to the cranial vertebra while the caudal vertebra was fixed to the test rig with the x-y-table allowing for translational movement orthogonal to the loading direction resulting in pure bending moments and pure shear forces. These kinematic boundary conditions were chosen, as further elaborated in the discussion.

**Biomechanical testing protocol**

Each specimen was tested load-controlled and displacement-controlled in the intact state and displacement-controlled after 7 (8) subsequent reduction steps, as described in the following section and in Fig. 2. After every reduction step, the segments were loaded in FE, LS, LB, AS, AR, and AC in the listed order. For each loading case, 5 preloading cycles were performed, before the relative movement between the cranial and caudal vertebral body was recorded during the sixth cycle. Motion of the two vertebrae was recorded with a motion capturing system (Atracsys Fusion Track 500, 10Hz record frequency, tracking accuracy 0.09 mm).

Loading was applied with a velocity of 1°/sec in FE and LB; 0.5°/s in AR, 0.5 mm/sec in AS, PS, and LS and 0.1 mm/sec in AC. The load-controlled experiments were conducted with ±7.5 Nm in FE, LB and AR, ±150 N in AS, PS, and LS and +400/-150 N in AC. These parameters conform to commonly used values in the literature, known as to be in a physiological range [10]. The following displacement-controlled testing steps were performed with the specific displacement magnitudes determined during load-controlled testing. Determination of the mounting positions (neutral zone) and the displacement magnitudes from the load-controlled step and assignment of these values back to the testing machine, was done completely automatized on the testXpert III software and a Matlab interface (Matlab R2019a, Mathworks Inc.) to prevent any operator errors. Initial preconditioning of each sample was performed in the intact state by repeating the load-controlled and two displacement-controlled loading cycles, until the difference between the 2 displacement-controlled measurements was below 0.25 Nm in rotational and below 5 N in translational loading, corresponding to an error of <3.3%. An average of 3–4 precondition cycles were required to achieve viscoelastic stable and reproducible conditions. During testing, specimens were kept moist by frequently spraying them with phosphate buffered saline.

**Description of the stepwise reduction**

All transection steps are illustrated in Fig. 3. (1) The ITL was cropped by cutting the transversal processes close to their origin with an oscillating saw. (2) The ISL&SSL were removed by cutting the spinous processes close to the dorsal aspect of the lamina, while care was taken not to harm the LF during this process. (3) The FJC on both sides were incised with a scalpel. Care was taken to incise all fibers of the joint capsule without harming the LF. (4) The two FJ were removed by cutting both the inferior and superior articular processes. Care was taken not to harm the LF. (5) To remove the LF, the inferior aspect of the superior lamina was cut horizontally, and the inferior lamina was removed by cutting it close to the pedicles. (6) The vertically oriented fibers of the ALL were identified and cut with a scalpel without harming the IVD by inserting the scalpel behind the ALL fibers and rotating the blade 90° anteriorly. (7) Analogously to the ALL, the vertically aligned fibers of the PLL were identified and cut without harming the IVD. (8) CT data
analysis, as well as visual inspection of the segments was conducted to identify any spondylophytes. If present, they were cut with care not to harm the adjoining section of the IVD.

**Data evaluation**

The load-deflection curves from the test machine during the sixth loading cycle were corrected with the 3D motion tracking data of the 2 vertebrae and used for analysis. The center line of the load-deflection hysteresis was calculated and fitted with a polynomial of the fifth order (Fig. 4A). Separation between positive and negative load application in load-deflection curves can be done by using the zero-crossing point, which is very robust in linear curves. However, in most cases the curves are S-shaped with a central lax zone and a progressive increase in stiffness during larger loads. In these cases, the zero-crossing point is extremely sensitive to minimal offset, while the minimal
slope is a more robust approach for separation. The logarithmic ratio between the maximal and minimal slope was used to quantify the S-shape of the curves (Fig. 4A) and in cases of this value being above 1.5, minimal slope was used, while in cases with values below 1.5, the zero-crossing point was used for separation (Fig. 4B). For LB, AR, and LS, pooled values between negative and positive load (left, right) were computed. The different load-deflection curves after resection were used to compute the contribution at different loading rates (Fig. 4C). The torsional pre-load in the sagittal plane was determined by analyzing the change in moment at the neutral position after each resection step in the flexion-extension configuration (Fig. 4D).

Specimens with severe IVD-degeneration (Pfirrmann 5 [12]) and segments with spondylolyphyes contributing to more than 15% in one of the tested loading directions were excluded resulting in a subgroup of 32 specimens (7 x L4L5, 9 x L3L4, 3 x L2L3, 7 x L1L2, 6 x T12L1) with little to moderate denegation (Table 1). Any spondylolyphyes-contribution was subtracted, and the contributions of the other structures was rescaled to 100% to allow for comparability.

Deflection amplitudes in AC were as little as 0.2 mm, leading to reduction curves in the hundredth mm range, which exceeded the measurement accuracy of the setup. It was therefore decided to not include this data in the here presented evaluation.

Results

The load apportionment of the passive spinal structures of 32 spinal segments with little to moderate degeneration is shown in Fig. 5. In flexion, the apportionments between the spinal structures is mostly shared between IVD (25%), PLL (16%), LF (22%), FJ with FJC (10%+14%), while the ISL&SSL contribution is small (6%) and the ALL and ITL contribute only minimally (<1% and <2%, respectively). In extension, the ALL (40%) and the IVD (36%) share almost three-fourth of the overall load, while the FJ and other passive structures contribute very little (<6%). During LB, the IVD is exposed to two-thirds of the overall load. Only the ALL provides relevant contribution in this load direction (15%), while all other structures provide only minimal support (<4%). In AR, the main load is transmitted over the FJ (37%), while the FJC (12%) and the IVD (32%) are the other major contributors. During LS, the IVD takes about 2/3 of the load (64%). The FJ contribute with 15%, while all other structures are measured with contributions smaller than 3%. Similarly, in AS the IVD takes about 2/3 of the load (67%) and the FJ contribute 14%. The highest contribution of the IVD is seen during PS (75%).

The contributing patterns of different levels show only small differences and no major trends can be observed (Fig. 5).

The load apportionment under different loading magnitudes (Fig. 6) is relatively constant, while certain notable trends can be observed: Increasing LF-contribution during application of smaller loads in flexion and increasing FJ-contribution during application of higher loads in AR, LS, and AS.

The ROM-normalized contribution of the passive structures during flexion-extension show small variability between samples (Fig. 7).

All passive structures of the lumbar spine are observed to provide some torsional preload in the sagittal plane (Fig. 8). The LF is the main contributor for preload in extension (0.25 Nm), while the IVD provides most preload in flexion (0.36 Nm). Conversion from torques to forces is not possible.

Discussion

Comprehensive knowledge of the load apportionment between the passive lumbar structures is of great value for preventive, prognostic, diagnostic and therapeutic approaches to spinal diseases. Even though numerous studies on spinal biomechanics have been published, many questions remain unanswered. This displacement-controlled transection study was performed to increase the understanding of the role and characteristics of the passive spinal structures.

Load directions

Specific and characteristic contribution patterns in the tested loading conditions are observed (Fig. 5): In flexion, it is conspicuous that multiple passive structures share the load, such as the PLL (16 %), the LF (22 %), the FJC (14 %) and the IVD (20 %). This is reasonable considering that these passive spinal structures are located dorsally. They support the autochthonous back muscles, which suffer from a limited lever arm for this loading direction. In extension, the segments are mainly stabilized through the ALL (40 %), while the contribution of the dorsally located facet joints is surprisingly small. The role of the FJ (37 %) and the FJC (12 %) seems to be mainly in AR, where they prevent excessive movement and therefore help protect against shearing of the neural structures. During LB, the IVD (67 %) is exposed to the majority of the load, only partially supported by the ALL (15 %), while the optimally located ITL (2 %) provides very limited support. In contrast to the autochthonous back muscles however, the abdominal wall muscles hold larger lever arms and could therefore be in a better position to provide stability. During shear loading, the IVD (64 %−75 %) is exposed to a large ratio of the stress (also seen by Tencer et Al. [2] and Cyron et al. [13]), while the FJ (14 %−15 %) provide progressive support in LS and AS (Fig. 6).

Principle of contribution conservation

A very notable observation in the here presented data is the “principle of contribution conservation”. A remarkably small functional variability in load-apportionment is observed when comparing different samples both from the same lumbar level as well as across different levels (Fig. 5). This is also reflected in the small standard deviations of the ROM-normalized ligament curves (Fig. 7). Similarly, large
Fig. 5. Contribution [%] (median (interquartile distance)) at 7.5 Nm moment as pie chart (left) and level depend (right) of intertransverse ligament (ITL), inter- and supraspinous ligament (ISL&SSL), facet joint capsule (FJC), facet joints (FJ), ligamentum flavum (LF), anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), intervertebral disc (IVD) in all loading directions.
Fig. 6. Load dependency: Contribution [%] (median) at 6 different loading steps (0.5, 1.5, 3, 4.5, 6, 7.5 Nm) of intertransverse ligament (ITL), inter- & supra-spinous ligament (ISL&SSL), facet joint capsule (FJC), facet joints (FJ), ligamentum flavum (LF), anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), intervertebral disc (IVD) in all loading directions.
consistent variations in the contribution patterns are observed at different loading magnitudes with only a few, well explainable trends (Fig. 6). This is remarkable, since the donor population is rather heterogeneous (Table 1), the ROM of the samples show relevant variability (e.g. FE-ROM variability between 5 - 13˚), the shape of the load-deflection curves range from almost linear to strongly S-shaped and since the different levels are anatomically different. The “principle of contribution conservation” indicates that the characteristics of the passive structures are accurately matched to one another and that they are adapted to the ROM of the specific segment. This harmonization prevents overloading of a single structure during any movement.

**Preload**

Another interesting observation is the presence of sagittal preload in the spinal structures (Fig. 8). While preload of the LF [14], the ALL [15], and the PLL [15] have been described in the healthy spine, the data of this study indicate that all structures are under pretension in the neutral position, which could provide some stability in the neutral zone. The preload could also ensure a certain hydrostatic environment in the unloaded condition (e.g. during sleep) which could be relevant for osmotic fluid and nutrition exchange and hence for cell viability. Furthermore, pretension can prevent ligament buckling during compressive forces, which is relevant for structures where buckling could result in neural compression.

**Anatomical structures**

Depending on their anatomical location and mechanical property, the passive structures provide specific contribution to the stability of the spine:

The ITL are intuitively expected to be of relevant importance during LB. Interestingly, neither in LB nor in any other loading direction, relevant contribution to the passive stability was observed. This is in agreement with the limited literature on this ligament [16]. We conclude that the ITL does not primarily provide passive stability but could play a role in proprioceptive sensing [17] and as muscle attachment point.

The ISL&SSL were observed to contribute only marginally to the passive stability of the spinal segments during the measured loading ranges. Small contribution during moderate loading was also observed by other authors [9,18–20]. This can be explained by the orientation of the collagen fibers.
some contribution during extension was observed (Figs. 5 and 7), which can be explained by some aspects of the LF being stretched during extension or by impingement of the LF between the laminae. The contribution of the LF during anterior shear is minimal and cannot explain the postulated relation between LF-degeneration and spondylolisthesis [24].

The ALL is the main contributor to the passive stability in extension, which can easily be explained by its anatomical location. Due to its anatomical extent from midline to quite far lateral [25], the contribution in LB can be understood. The measured preload in the neutral position is in agreement with the literature [15] and is required to counteract the preload of the dorsal structures.

The PLL contributes mainly in flexion, while its contribution in all other tested loading conditions is minimal. The small preload of the PLL could help prevent buckling into the spinal canal during compression and extension.

The IVD is of great importance in all loading directions. While the IVD is shielded from the majority of the load in flexion, extension and AR, it is exposed to a high percentage of the loads in LB, AS, PS and LS.

**Clinical implications**

With the IVD being of great importance to the passive stability of the lumbar spine, it is also exposed to a relevant proportion of the resulting stresses. Since IVD degeneration is responsible for relevant morbidity as well as socioeconomic costs [26], the role of the muscles as active stabilizers and the relevance of posture and movement optimization becomes evident.

During surgery of the lumbar spine, different anatomical structures are regularly incised or dissected. In midline decompression with laminotomy, the SSL, the ISL and the LF are resected, which would result in the loss of about 28% of the passive stabilizers in flexion. Unilateral microsurgical decompression with resection of about 50% of the LF would result in 11% reduction of the passive stability in flexion. At the present, it is still unclear whether less invasive decompression such as ISL&SSL sparing techniques could reduce postoperative back pain or relevant instability [27]. Good clinical results of these surgeries [28] prove the ability of the human body to compensate for the loss of these structures, most probably by stabilizing muscle function. However, in certain cases, the destabilizing effect of decompression surgeries are not well tolerated resulting in pain and malfunction [29–31]. It is notable that postoperative instability is mostly defined as olisthesis [29–32], which implies loss of stability in anterior shear. According to the here presented results, the IVD, FJ and the FJC are the most important stabilizing structures in this regard and therefore their preservation during decompressive surgery should be guaranteed, whereas resection of the ISL&SSL should not generate large instability during mid-range movements. However, the role of the ISL&SSL as an end

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Fig. 8. Preload [Nm] of intertransverse ligament (ITL), inter- and supraspinous ligament (ISL&SSL), facet joint capsule (FJC), facet joints (FJ), ligamentum flavum (LF), anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), intervertebral disc (IVD) in flexion-extension. (*see limitations)
stop in flexion remains unclear and their resection could possibly generate vulnerability to extensive flexion loads.

**Academic implications**

From a computational simulative perspective, the “principle of contribution conservation” is of paramount importance. Accurate tuning of the parameters is required to prevent unrealistic load apportionment in numerical and multi-body dynamic simulations due to generalized material properties. Tuning must be done anatomy- and segment-specific.

**Limitations**

The semi-constrained setup used for this study provides excellent control over the rotational movements since the rotation around the z-axis is actively controlled and rotational movements around the x- and y-axis are constrained. The translational freedom provided by the x-y-stage results in pure moment around the z-axis and pure compressive forces along the z-axis. As a consequence, however, coupled motion in the x-y plane are possible, allowing for small changes in movement patterns after each resection step, which could result in a slight over- or underestimation of the contribution due to the resection sequence. Despite this limitation, the semi-constrained setup was favored over a completely constrained setup to allow for pure moment and pure force generation, which we believe is necessary to resemble the in-vivo kinematics adequately [33]. A completely un-constrained test setup (as proposed by other authors for biomechanical testing of the spine [34]) allows for complex, three-dimensional coupled motions, which can be of great relevance in certain biomechanical experiments but introduce more uncontrolled parameters. With accurate and unbiased recordings of the contribution patterns being the primary aim of this study, the characteristics of semi-constrained testing was favored to testing in a completely un-constrained setup.

The described method for separation between positive and negative loading (e.g. flexion/extension) was considered to be most ideal for this purpose, however it could potentially introduce some bias. Stepwise reduction of the anatomical structures was performed with outermost prudence, however certain limitations could not be prevented. The anteromedial aspect of the FJC could not be reached in certain cases, resulting in remnants of the FJC being resected with the FJ. Also, injury to the LF during FJ-resection is not completely avoidable. These limitations can explain the contribution of the FJ during flexion (Fig. 5) and the preload (Fig. 8) measured at the FJ, which cannot arise from the joint surfaces alone. Since all testing was performed on spinal segments and not on complete spinal columns, dissection of ligaments covering multiple levels like the ALL, PLL and SSL could lead to underestimation of their contribution [35]. Conclusions can only be drawn on isolated movements, since no combined movements were tested.

**Conclusions**

The present work is the first to quantify the contribution of the passive spinal structures in a load-controlled manner during physiological loading. The IVD takes the main load in LB and absorbs shear loading while the FJ&FJC stabilize AR. The ALL resists extension while LF and PLL stabilize flexion. With the newly observed “principle of contribution conservation”, isolated analysis of spinal structures seems to be insufficient, since the characteristics of the passive structures are well matched to one another. While the adaptation of the passive structures to each other is evident, the cause for the specific segmental stability remains unclear and is an interesting field of research for further studies.

**Conflict of interest statement**

None to declare.

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