3rd International Workshop on

Spine Loading and Deformation

Program and Abstract Book

Berlin 4 - 6 July 2019

JULIUS WOLFF INSTITUT

CHARITÉ
UNIVERSITÄTSMEDIZIN BERLIN

Thursday, July 4th	Friday, July 5th	Saturday, July 6th
10:00–12:45 **Registration** Coffee & Snack	8:00–10:00 **Session 3** Motion Segments: Load Sharing	8:00–9:35 **Session 7** Spinal Loads – Computational Models
12:45–13:00 **Welcome**	10:00–10:30 **Coffee Break**	9:35–10:30 **Coffee to Go –** **Poster Session**
13:00–14:00 **Session 1** Intervertebral Disc – Tissue Mechanics	10:30–12:30 **Session 4** Lumbar Spine I: Shape and Kinematics	10:30–12:45 **Session 8** Trunk Stabilization and Control
14:00–14:20 **Coffee Break**	12:30–13:30 **Lunch Break**	12:45–13:00 **Final Words**
14:20–15:50 **Session 1** (cont.) Intervertebral Disc – Tissue Mechanics	13:30–15:30 **Session 5** Lumbar Spine II: Loads and Kinematics – Injury/Degeneration/Pain	13:00–14:00 **Lunch**
15:50–16:20 **Coffee Break**	15:30–16:00 **Coffee Break**	
16:20–18:00 **Session 2** Debate on Mechanical Load – Injury/Degeneration/Pain	16:00–18:00 **Session 6** Spinal Loads – In-Vivo Measurements and Modeling	
18:00 **Happy Hour**	19:00 **Dinner**	

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Venue

Julius Wolff Institute
Charité - Campus Virchow Klinikum
Institutsgebäude Süd
Föhrer Straße 15, 13353 Berlin

Date

4 - 6 July 2019

Organizers

Hendrik Schmidt
Julius Wolff Institute – Charité
Berlin, Germany

Saeed A. Shirazi-Adl
École Polytechnique
Montréal, Canada

Idsart Kingma
Vrije Universiteit Amsterdam
The Netherlands

Coordinator

Friedmar Graichen friedmar.graichen@charite.de

Secretary

Barbara Schiller barbara.schiller@charite.de

Welcome Note of the Congress Participants

Dear colleagues and friends,

We are pleased to welcome you to the **3rd International Workshop on Spine Loading and Deformation** which will be held in Berlin during 4-6 July 2019.

Mechanical loading and deformation in the human spine during diurnal activities are recognized to play a major role in the etiology of back disorders and pain. A comprehensive knowledge of these loads/deformations and associated interaction with the tissue response is a basic prerequisite for effective risk prevention and assessment in the workplace and in sports and rehabilitation, for proper management of various disorders, and for realistic preclinical testing of spinal implants.

However despite considerable advancement and numerous investigations, many crucial issues remain yet unresolved. In vivo, in vitro and computational model studies in spine biomechanics as well as connected disciplines are all necessary for tangible progress in this field.

This workshop on Spine Loading and Deformation aims to bring world-wide researchers active in this field together in order to share and discuss their recent works on related areas and explore the potentials of their findings. The research topics cover trunk loads and motions (imaging, sensors and video camera) measurements/predictions during sports, occupational tasks, perturbations, etc. with focus on the lumbar and thoracic spines.

We cordially welcome you all to this **3rd International Workshop on Spine Loading and Deformation** and wish you again an enriching scientific meeting and a pleasant stay in Berlin.

Yours,
Hendrik Schmidt
Idsart Kingma
Saeed A. Shirazi-Adl

10:00-12:45 Foyer	**Registration / Coffee & Snack**
12:45-13:10 Lecture Hall	**Welcome and Workshop Opening Remarks** Hendrik Schmidt, Saeed Shirazi-Adl & Idsart Kingma
13:10-15:50 Lecture Hall	**Session 1: Intervertebral Disc – Tissue Mechanics** Moderators: Peter-Paul Vergroesen, Grace O'Connell
13:10	Time Dependent Behavior of Pressure and Height of a Loaded Intervertebral Disc are Inconsistent Kaj Emanuel (Amsterdam, The Netherlands)
13:25	Annulus Fibrosus Hydration Affects Rate-Dependent Failure Mechanics In Tension Grace O'Connell (Berkeley, USA)
13:40	On the Modeling of Human Disc Annulus Fibrosus: Elastic, Yield and Failure Responses Farshid Ghezelbash (Montréal, Canada)
13:55	Internal Deformations in Human Intervertebral Discs: a 9.4T MRI Study Nicolas Newell (London, UK)
14:10-14:35	**Coffee Break**
14:35	In-Vitro Perspective into Micro-structural Degeneration of the Intervertebral Disc: a Biomechanical Approach David Rivera Tapia (Exeter, UK)
14:50	Collagen Fiber Bundles Disintegration during Pull-out from the Endplate Magdalena Wojtków (Wrocław, Poland)
15:05	Voltage-Gated Ion Channels in Intervertebral Disc Mechanotransduction Philip Poillot (Limerick, Ireland)
15:20	Open Discussion
15:50-16:20	**Coffee Break**

16:20-18:00 Lecture Hall	**Session 2: Debate on Mechanical Load – Injury/Degeneration/Pain** Moderators: Saeed Shirazi-Adl, Hendrik Schmidt
16:20	Opening Presentation: Low Back Pain Paradox Saeed Shirazi-Adl (Montréal, Canada)
16:35	Intervertebral Disc Degeneration from a Biomechanical Point of View: What do we need to fix? Peter-Paul Vergroesen (Amsterdam, The Netherlands)
16:50	Spine Postures, Physical Exposure, and Back Pain: a Systematic Review of Systematic Reviews Daniel Belavy (Burwood, Australia)
17:05	Interactions between Genetics and Loading in Development of Disc Degeneration and Low Back Pain – a Review Jill Urban (Oxford, UK)
17:20	Open Discussion
18:00	**Happy Hour** Beer, Pretzel, Snacks and Live Jazz Music

08:00-10:00 Lecture Hall	**Session 3: Motion Segments: Load Sharing** Moderators: Babak Bazrgari, William Anderst
08:00	Review of Load-sharing in Intact, Transected, Degenerate and Surgically Altered Passive Human Lumbar Spines Marwan El-Rich (Abu Dhabi, UAE)
08:15	Relationship between Intervertebral Disc and Facet Joint Degeneration: a Probabilistic Finite Element Model Study Maxim Bashkuev (Berlin, Germany)
08:30	Lumbar Spinal Ligament Characteristics extracted from Stepwise Reduction Experiments allow for Precise Modeling Nicolas Damm (Koblenz, Germany)
08:45	Detailed Full-filed Analysis of the Ventral Lumbar Spine: Insights on the Biomechanical Role of the Anterior Longitudinal Ligament Luigi La Barbera (Montréal, Canada)
09:00	Effects of Lumbar Lordosis on Mechanical Response of Post-operative Lumbar Spine – Personalized Parametric Finite Element Simulations Mohammad Nikkhoo (Tehran, Iran)
09:15	Biomechanical Properties in Motion of Lumbar Spines with Degenerative Scoliosis Idsart Kingma (Amsterdam, The Netherlands)
09:30	Open Discussion
10:00-10:30	**Coffee Break**
10:30-12:30 Lecture Hall	**Session 4: Lumbar Spine I: Shape and Kinematics** Moderators: André Plamondon, William Marras
10:30	Review Article on Spine Kinematics of Quadrupeds and Bipeds Sandra Reitmaier (Berlin, Germany)
10:45	Dynamic Interactions between Lumbar Intervertebral Motion Segments during Forward Bending Alexander Breen (Bournemouth, UK)
11:00	Sex-dependent Difference in Lumbo-Pelvic Coordination for Different Lifting Tasks Fumin Pan (Berlin, Germany)

11:15	A Novel Model and Experimental Validation Demonstrate the Large Contribution of Passive Muscle to Spine Flexion Relaxation Stephen Brown (Guelph, Canada)
11:30	Calculating the Three-dimensional Vertebral Orientation from a Planar Radiograph: is it feasible? Fabio Galbusera (Milan, Italy)
11:45	Which Landmark is Best Suited to Assess the Thoracic Orientation? Thomas Zander (Berlin, Germany)
12:00	Open Discussion
12:30-13:30	**Lunch Break**
13:30-15:30 Lecture Hall	**Session 5: Lumbar Spine II: Loads and Kinematics – Injury/Degeneration/Pain** Moderators: Navid Arjmand, Daniel Belavy
13:30	In-Vivo Hip and Lumbar Spine Implant Loads during Activities in Forward Bent Postures Philipp Damm (Berlin, Germany)
13:45	Bottom-up Versus Top-down L5/S1 Moment Estimation during Manual Lifting using an Ambulatory Measurement System Gert Faber (Amsterdam, The Netherlands)
14:00	A Prospective Study of Lumbo-pelvic Coordination in Patients with Non-chronic Low Back Pain Babak Bazrgari (Lexington, USA)
14:15	Patient-Specific Changes in Adjacent Segment Kinematics After Lumbar Decompression and Fusion William Anderst (Pittsburgh, USA)
14:30	The Impact of Curve Severity on the Pelvic Kinematic and Erector Spinae and Gluteusmedius Muscles Activity during Gait in Patients with Adolescent Idiopathic Scoliosis Shirin Yazdani (Tabriz, Iran)
14:45	Automatic Generation of Patient-Specific FE Models of the Lumbar Spine Sebastiano Caprara (Zurich, Switzerland)
15:00	Open Discussion
15:30-16:00	**Coffee Break**

16:00-18:00 Lecture Hall	**Session 6: Spinal Loads – In-Vivo Measurements and Modeling** Moderators: Idsart Kingma, Ameet Aiyangar
16:00	Effect of a Passive Exoskeleton on Mechanical Loading during Dynamic Lifting Axel Koopman (Heemskerk, The Netherlands)
16:15	Sex-Dependant Estimation of Spinal Loads during Static Manual Material Handling Activities- combined In Vivo and In Silico Analyses Ali Firouzabadi (Berlin, Germany)
16:30	Subject-specific Regression Equations to Estimate Spinal Loads in Asymmetric Static Lifting André Plamondon (Montréal, Canada)
16:45	Sensitivity of Musculoskeletal Model-based Lumbar Spinal Loading Estimates to Type of Kinematic Input and Passive Stiffness Properties Ameet Aiyangar (Duebendorf, Switzerland)
17:00	Assessment of Spine Loading via a 2 Muscle Model vs. 10 Muscle Model during One vs. Two Handed Lifting Tasks William Marras (Powell, USA)
17:15	Estimating Lumbar Passive Stiffness Behaviour from Subject-specific Finite Element Models and In Vivo 6DOF Kinematics Christian Affolter (Duebendorf, Switzerland)
17:30	Open Discussion
19:00	**Departure of the Bus to the Restaurant**
19:30	**Social Event: Dinner**

08:00-09:35 Lecture Hall	**Session 7: Spinal Loads – Computational Models** Moderators: Dominika Ignasiak, Fabio Galbusera
08:00	A Novel Method for Prediction of Postoperative Global Sagittal Alignment based on Full-Body Musculoskeletal Modeling and Posture Optimization Dominika Ignasiak (San Diego, USA)
08:15	Predicting Intervertebral Disc Loading and Trunk Muscle Activity in Healthy Adolescents using Musculoskeletal Full-Body Models Stefan Schmid (Boston, USA)
08:30	Coupled Artificial Neural Networks to Predict Whole Body Posture, Lumbosacral Moments, Trunk Muscle Forces, and Lumbar Disc Loads during Three-dimensional Material Handling Activities Navid Arjmand (Tehran, Iran)
08:45	Influence of Seat Parameters on Computationally Predicted Spine Loading Xuguang Wang (Bron, France)
09:00	Statistical Shape Model Predicted Alignments and Musculoskeletal Simulation in Surgical Planning Jess Snedeker (Zurich, Switzerland)
09:15	Open Discussion
09:35-10:30 Foyer	**Coffee to Go - Poster Session** All Poster Presenters of P1 - P13
10:30-12:45 Lecture Hall	**Session 8: Trunk Stabilization and Control** Moderators: Jaap van Dieën, Christian Larivière
10:30	Trunk Stabilization in Patients with Low-back Pain and Healthy Controls Jaap van Dieën (Amsterdam, The Netherlands)
10:45	Indication of Diagnostic Criteria for Proprioception Disorders between Non-specific Low Back Pain Patients and Healthy People based on Analysis of Linear and Nonlinear Parameters of Center of Pressure and Trunk Stability Mohamad Parnianpour (Tehran, Iran)

11:00	Sudden Gait Perturbations elicit Sex-specific Neuromuscular Trunk Responses in Persons with Low Back Pain Juliane Mueller (Trier, Germany)
11:15	Can Trunk Postural Control During Unstable Sitting be considered a Proxy Measure of Dynamic Lumbar Stability? Christian Larivière (Montréal, Canada)
11:30	Biomechanics of Intra-Abdominal Pressure in Spine Stiffening and Loading - A Systematic Review of In Vivo and Modeling Studies Navid Arjmand (Tehran, Iran)
11:45	Real-time Feedback to Reduce Lower Back Moment while Lifting a Box: a Proof-Of-Concept-Study Michiel Punt (Amsterdam, The Netherlands)
12:00	Reducing the Number of Input Variables Required to Control an Active Exoskeleton Ali Tabasi (Amsterdam, The Netherlands)
12:15	Open Discussion
12:45-13:00	**Final Words**
13:00-14:00	**Lunch**

09:35-10:30 **Coffee to Go – Poster Session**
Foyer Presenters are prepared for discussions in front of their posters

P1 Influence of the Facet Joints on the Mechanical Behaviour of the Intervertebral Disc: the Numerical and Experimental Analysis
Małgorzata Żak (Wrocław, Poland)

P2 Beyond Preload - The Replication of Six-axis In-Vivo Load Data using a Spine Simulator
Timothy Holsgrove (Exeter, UK)

P3 Two-level Fusion Versus Topping-off Technology based on Coflex in the Treatment of Lumber Degenerative Disease: a Biomechanical Effect Comparison
Xiang-Yao Sun (Beijing, China)

P4 Sublaminar Tape as Alternative and Addition to Pedicle Screws in Spinal Surgery
Remco Doodkorte (Maastricht, The Netherlands)

P5 Effects of the Nucleus Migration during Forward Flexion on the Biomechanics of the L4-5 Functional Spinal Unit
Marwan El-Rich (Abu Dhabi, UAE)

P6 Sensitivity of Musculoskeletal Model Predictions in Neutral Standing and Forward Flexion Postures to Center of Rotation Location
Marwan El-Rich (Abu Dhabi, UAE)

P7 A new Method for Validation of an Individual Forward Dynamics Model of the Lumbar Spine
Nicolas Damm (Koblenz, Germany)

P8 Single Rigid Segment versus Multi-segmental Approach for the Analysis of the Lumbar Spine in Low Back Pain
Enrica Papi (London, UK)

P9 The Workload on One's Low Back during Dish-washing
Han Zhang (Shanghai, China)

P10 Smart Rotational Spine Protector (RSP) for Sport and Rehabilitation
Dietmar Rafolt (Vienna, Austria)

P11 Biomechanical Evaluation of PEEK Semi-Rigid Fixation Subject to Static and Cyclic Loading
Kinda Khalaf (Abu Dhabi, UAE)

P12 Using SHARIF-HMIS Inertial Sensor for Measurement and Comparison of Kinematic Parameters in 3 Subgroups of STarT Back Screening Tool in Patients with Nonspecific Low Back Pain
Mohamad Parnianpour (Tehran, Iran)

P13 Risk for Fatigue-related Degeneration of the L5-S1 Disc among Persons with vs. without Unilateral Lower Limb Amputation
Babak Bazrgari (Lexington, USA)

Abstracts Podium

Time Dependent Behavior of Pressure and Height of a Loaded Intervertebral Disc are Inconsistent

van der Veen AJ[a], Emanuel KS[a], van Dieën JH[b]

[a]*VU University Medical Center, Amsterdam, The Netherlands*
[b]*VU University Amsterdam, Amsterdam, The Netherlands*

Sustained loading on the intervertebral disc leads to loss of disc height. The generally accepted explanation for this is that the disc loses height due to an imbalance between the external load on the disc and the osmotic pressure in the disc. Water is expelled from the disc until the osmotic attraction reaches an equilibrium with the applied load. In this study, we compared the time course of loss of disc height and pressure.

Fourteen caprine lumbar discs were tested in a saline bath. Vertebral bodies were cut-off close to the endplate. A pressure needle was inserted in the nucleus. After preloading (10N, 6hours) an axial load of 150N (18hours) was applied to the disc. A double Kelvin-Voigt model, which represents summation of a fast and a slow decay function, was fitted to the data. Time constants for of change of pressure and disc height were calculated. Paired t-tests were used to compare both time constants.

The time constant of the slow decay function was larger for disc height than for nucleus pressure ($p=0.0006$). The difference in time constants of both fast decay functions was not significant ($p=0.7$).

We found a difference between the time constants of the change of the disc height and of the nucleus pressure. The discs reached an equilibrium between internal and external pressure well before the change of disc height came to a stop. This indicates that the change of disc height depends on more than pressure equilibrium alone. Likely, viscoelastic properties of the annulus fibrosis play an important role as well.

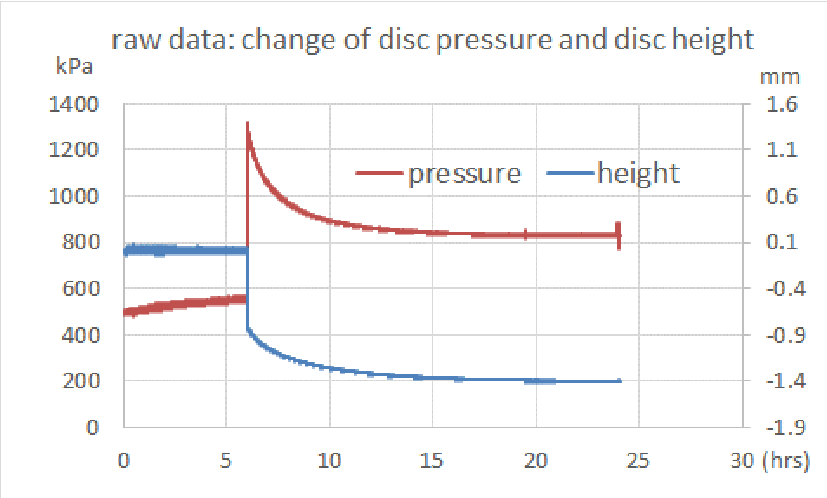

Fig. 1: Typical example of the change in disc height and disc pressure in time. The first test phase (6 hours) is preloading the disc, the second test phase is applying the load, which is shown as a vertical line in the graph and finally the creep phase in which the applied load is maintained at a constant level leading to a decrease in both the nucleus pressure and disc height in time. Time constants of decrease in height and pressure have been calculated during this phase. Time constant of the slow decay of disc height is larger than the time constant of the change of pressure.

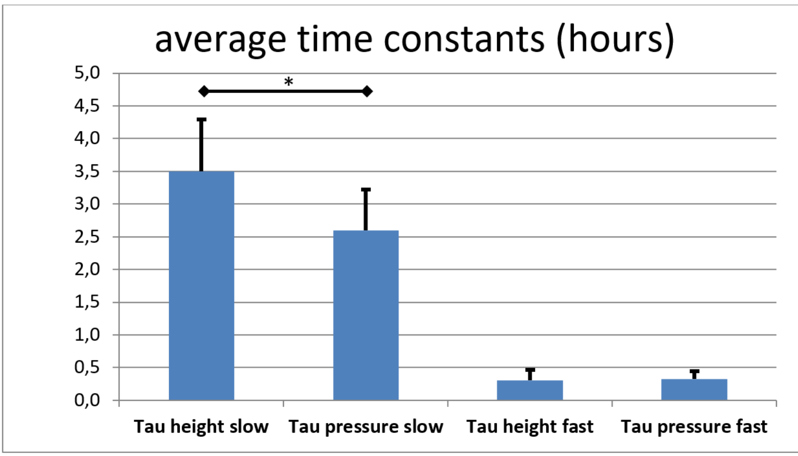

Fig. 2: Time constants of both decay functions. The slow decay function is significantly different. Fast decay function is not.

Annulus Fibrosus Hydration Affects Rate-Dependent Failure Mechanics In Tension

Werbner B[a], Spack K[a], O'Connell GD[a,b]

[a]*Dept. of Mechanical Engineering, University of California, Berkeley, CA, USA*
[b]*Dept. of Orthopaedic Surgery, University of California, San Francisco, CA, USA*

The high water content of the intervertebral disc is essential to its load bearing function and viscoelastic mechanical behavior. One of the primary biochemical changes associated with disc degeneration is the loss of proteoglycans, which is associated with tissue dehydration. While previous studies have reported the effects of in vivo degeneration on annulus fibrosus (AF) failure mechanics, the independent role of water content remains unclear, as does the tissue's rate-dependent failure response. Thus, our first objective was to determine the effect of loading rate on AF failure properties in tension; the second objective was to quantify the effects of water content on the failure properties.

Water content was altered through enzymatic digestion of glycosaminoglycans (GAGs) and through osmotic loading. AF specimens from bovine caudal discs were tested monotonically to failure at a rate of 0.05mm/min or 50mm/min. Increased loading rate resulted in a ~50% increase in linear-region modulus, failure stress, and strain energy across all treatment groups (Fig. 1). Lower GAG and water contents resulted in decreases in modulus, strength, and strain energy; however, these differences were only observed at the low loading rate ($p<0.05$; no changes at high rate). Osmotic loading was used to isolate the effect of hydration, separate from GAG composition, resulting in similar decreases in water content, modulus, and strain energy (Fig. 2). These results suggest that tissue hydration is essential for maintaining bulk tissue stiffness and capacity for energy absorption, rather than strength. These findings also suggest that GAGs may contribute to tissue strength, independent from its role in mediating water content, possibly through fiber-matrix interactions, which will be the focus of future work. In conclusion, this study provides new insights into the structure-function relationship between AF water content and tensile mechanics.

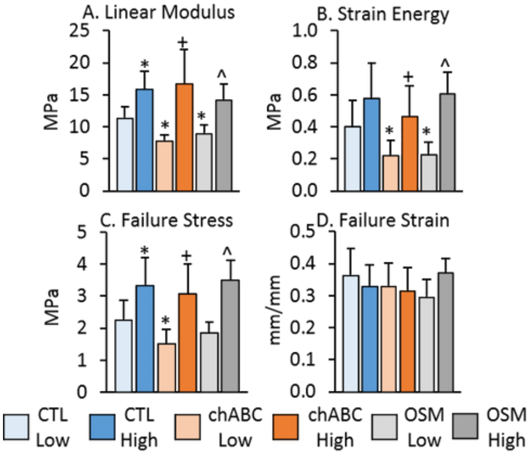

Fig. 1: Summary of mechanical properties in uniaxial tension (n = 11 per group): A) linear-region modulus, B) strain energy to the point of failure, C) failure stress, and D) failure strain. * denotes $p < 0.05$ vs CTL-Low, + denotes $p < 0.05$ vs chABC-Low, ^ denotes $p < 0.05$ vs OSM-Low in post-hoc analysis.

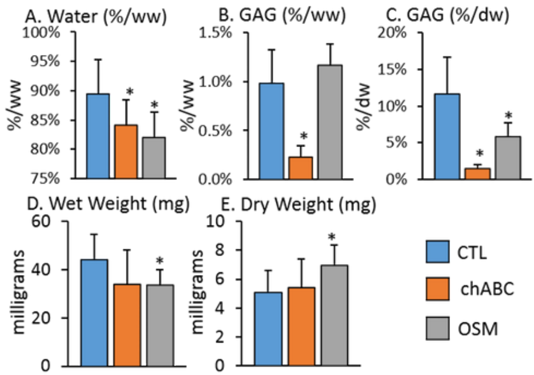

Fig. 2: Summary of biochemical data (n = 22 per group): A) water content normalized by swollen weight (ww), B) GAG content normalized by swollen weight, C) GAG content normalized by dry weight (dw), and D) swollen and E) dry weights from biopsy punch. Biochemical data from low- and high-rate loading groups were pooled. * denotes $p < 0.05$ vs CTL in post-hoc analysis.

On the Modeling of Human Disc Annulus Fibrosus: Elastic, Yield and Failure Responses

Ghezelbash F[a], Baghani M[b], Shirazi-Adl A[a], Eskandari A[c]

[a]*Polytechnique Montréal, Québec, Canada*
[b]*University of Tehran, Tehran, Iran*
[c]*Sharif University of Technology, Tehran, Iran*

Annulus fibrosus (AF) plays an important role in spine biomechanics. Finite element studies have developed numerous models with varying assumptions. Limited model studies have investigated experimental and in vivo observations of the damage, permanent deformation and rupture in AF tissue. The aim of this study is two-fold: 1- developing a non-homogenous AF model incorporating permanent deformation and damage in both collagen fibers and matrix, and 2- critically evaluating the relative performance of our model and other reported models in comparison with existing tissue-level (uniaxial and biaxial tests) and whole disc studies.

We used compressible neo-Hookean strain energy function to simulate the elastic response of the AF matrix. To capture the permanent deformation, an inelastic dissipation energy function similar to the neo-Hookean model was considered [1]. Collagen fibers were simulated using uniaxial truss elements with nonlinear elastic response followed by a nonlinear isotropic strain-hardening [2]. For comparison, we selected five additional AF models from the literature to critically evaluate the performance of existing AF models with available tissue-level experiments in uniaxial (circumferential, radial and axial directions) and biaxial tests. After calibration in tissue-level tests, the proposed model was incorporated in a lumbar disc-body unit finite element model to evaluate its performance in a whole disc model under compression.

Our proposed model demonstrated overall satisfactory agreement with tissue-level experiments in uni- and bi-axial tests; other AF models, however, showed discrepancies at least in one of the tissue-level tests (Fig. 1). The model also accurately predicted the failure response of the tissue in different testing directions (Fig. 2). The collagen fiber content substantially affected the failure response greater than the fiber angle (Fig. 2). The finite element disc model reasonably well predicted compression-displacement response with some permanent deformations noted in AF matrix at compression <4000 N.

Validation of an AF model should be performed in both uniaxial (different directions) and biaxial tests since a relative agreement in one test does not guarantee the validity in remaining tests. Proper considerations of the yield-failure in fibers and permanent deformation in matrix allow for the accurate prediction of post-yield and failure responses of AF tissue.

[1] Fereidoonnezhad et al. (2016) J Mech Behav Biomed Mat, 61:600-616
[2] Shirazi-Adl et al. (1986) Spine, 11:914-927

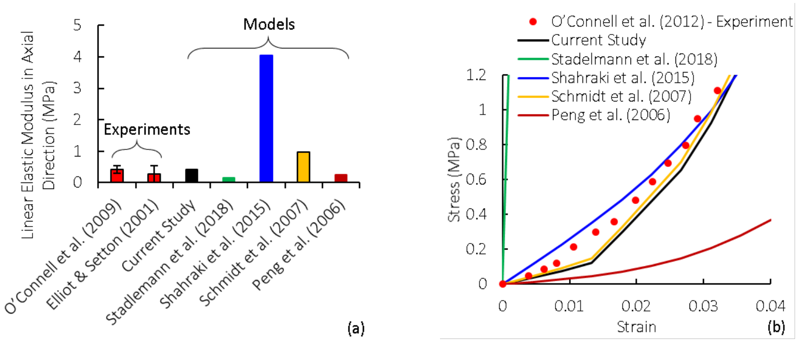

Fig. 1: Model predictions versus measurements; (a) linear elastic modulus in uniaxial tensile tests in axial direction, and (b) nominal stress-strain curves in circumferential direction under equi-biaxial tensile tests (curves in axial direction look similar but are not shown). Collagen fiber content (η) and fiber orientation (φ) were set at 15% and 35°.

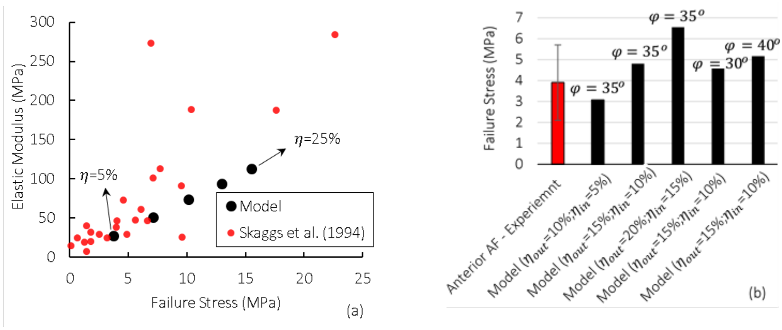

Fig. 2: (a) Failure stress versus elastic modulus (at 75% of failure strain) of AF samples under uniaxial tension in fibers direction for the model (with various collagen contents - η: 5, 10, 15, 20 and 25%) as well as measurements (Skaggs et al., 1994). (b) Failure stress predicted in the model (for different fiber orientations (φ) and collagen contents varying linearly from outer (η_{out}) layers to inner layers (η_{in}) since in experiments bone-tissue-bone samples were quite thick ~5 mm) versus measurements (Green et al., 1993) under uniaxial tension along axial direction.

Internal Deformations in Human Intervertebral Discs: a 9.4T MRI Study

Tavana S[a], Prior J[b], Baxan N[c], Hansen U[a], Masouros SD[b], Freedman B[d], Newell N[a]

[a]*Dept. of Mechanical Engineering, Imperial College London, London, UK*
[b]*Dept. Bioengineering, Imperial College London, London, UK*
[c]*Biomedical Imaging Centre, Dept. of Medicine, Imperial College London, London, UK*
[d]*Mayo Clinic, Rochester, MN, USA*

Back pain will be experienced by 70-85% of all people at some point in their lives [1], and is linked with intervertebral disc (IVD) degeneration [2]. However, few studies have attempted to quantify changes in the internal deformations within IVDs as they degenerate. Recent advances in MRI technology provide the opportunity to observe 3D deformations within intact IVDs in unprecedented detail. The aim of this study was to quantify human IVD deformations under axial compression using 9.4T MRIs.

Two degenerate, and two non-degenerate human vertebral body–IVD–vertebral body specimens (L4-L5) were used for this study. Specimens were aligned with the transverse plane of the disc parallel to the base of the mounting pots of a custom made compression rig, and fixed in place using polymethyl methacrylate (PMMA). MRIs were acquired before, and after 2mm of compression (Fig. 1). Digital Volume Correlation (DVC) was used to calculate the 3D strains within the IVDs.

High lateral strains were seen in the AF regions of the non-degenerate discs, while high lateral strains were seen in both the AF and the NP of the degenerate discs, particularly close to the endplates (Fig. 2).

This is the first study to use high field MRI to obtain images with in plane resolution as high as (90 x 90) μm^2 to investigate internal deformations within degenerate and non-degenerate human discs. The 3D strain maps are useful for designers of partial or total disc replacement technologies who aim to restore the mechanical behaviour of degenerate discs back to their non-degenerate state. Future investigations into 3D strains under different modes of loading will inform physical activities to mitigate high IVD strains, and the optimisation of rigid instrumentation for fusion surgery by ensuring that modes of loading associated with the highest IVD strains are minimised.

[1] Andersson GB. Lancet 1999;354:581–5.
[2] Luoma K et al. Spine (Phila Pa 1976) 2000;25:487–92.
[3] Pfirrmann CW. Spine (Phila Pa 1976) 26:1873-1878, 2001.

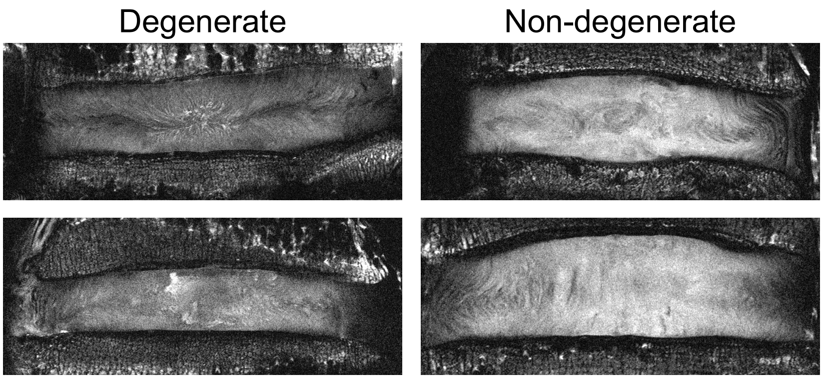

Fig.1: Middle coronal plane slices of MRIs (T2-weighted RARE) of the four unloaded human specimens. Images on the left are of degenerate specimens (Pfirrmann grade ≥ 3), while images on the right are non-degenerate (Pfirrmann grade ≤ 2) [3].

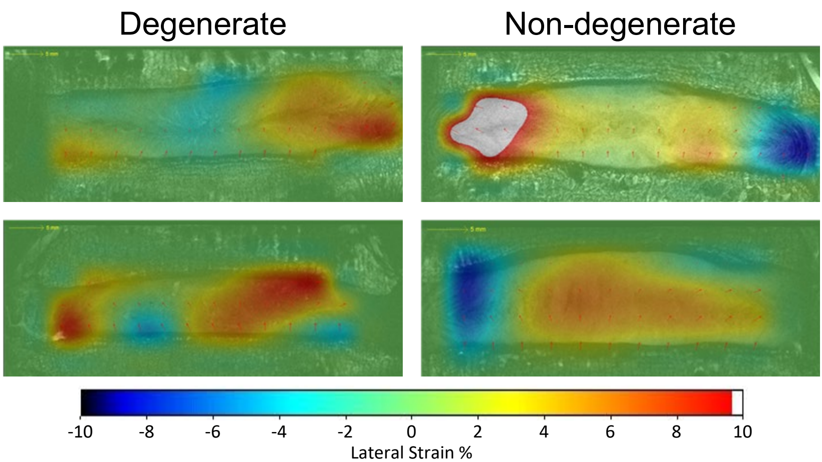

Fig.2: Lateral strain maps from the middle coronal plane of each specimen under 2mm of compressive displacement. Negative strains represent compression, positive strain represents tension.

In-Vitro Perspective into Micro-structural Degeneration of the Intervertebral Disc: a Biomechanical Approach

Rivera Tapia ED[a], Meakin JR[b], Holsgrove TP[a]

[a]*Dept. of Engineering, University of Exeter, Exeter, UK*
[b]*Dept. of Physics, University of Exeter, Exeter, UK*

Intervertebral disc degeneration is a deteriorating disorder commonly associated with lumbar back pain, leading to both high treatment costs and short- and long-term physical impairments [1, 2]. Previous studies have initiated disc damage through uniaxial static loading [3], vibration in subcritical conditions [4] or puncturing [5]. The outcomes have contributed to understanding some of the effects of degenerative mechanisms, however, mechano-optical investigations will provide a greater understanding, increasing our ability to translate this research into clinical benefit. This study considered sinusoidal overloading to initiate disc degeneration using an in-vitro bovine tail model. Specimens were tested either a control (CTL) or overload (OVL) group. The loading regime for both groups comprised 8 hours preconditioning load (0.5 MPa) [6], followed by 6 compressive periods distributed in 2 hours of sinusoidal loading (1 Hz, 0.2 – 0.8 MPa) and 1 hour of static recovery (0.2 MPa). The groups differentiated in the 4th sinusoidal compressive period of the OVL group (2 hours, 5 Hz, 1.0 – 2.6 MPa) (Fig. 1). The net disc height (NDH) was consistent after all 6 periods (20 – 30 µm) in the CTL cases and for the first 3 periods in the OVL group. The NDH in the OVL group following overloading in the 4th period remained positive, (Fig. 2a), suggesting further periods are necessary to stabilise the NDH, and that overloading prevents disc height recovery from occurring. The stiffness was not significantly altered ($p > 0.800$) between groups at any period except during the overload ($p = 0.01$) (Fig. 2b). Similar overloading regimes have led to microstructural damage [4]; this study demonstrates such overloading does not immediately alter mechanical properties at the organ-scale. Multi-photon microscopy of the tested specimens will provide a valuable link between the micro- and macro-structure of the disc under both normal and overloading conditions.

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[2] Belfiore P, Scaletti A, Frau A, Ripani M, Spica VR, Liguori G. Economic aspects and managerial implications of the new technology in the treatment of low back pain. Technol Health Care 2018;1:1–10. doi:10.3233/THC-181311.
[3] Adams MA, Freeman BJC, Morrison HP, Nelson IW, Dolan P. Mechanical Initiation of Intervertebral Disc Degeneration. Spine (Phila Pa 1976) 2000;25:1625–36. doi:10.1097/00007632-200007010-00005.
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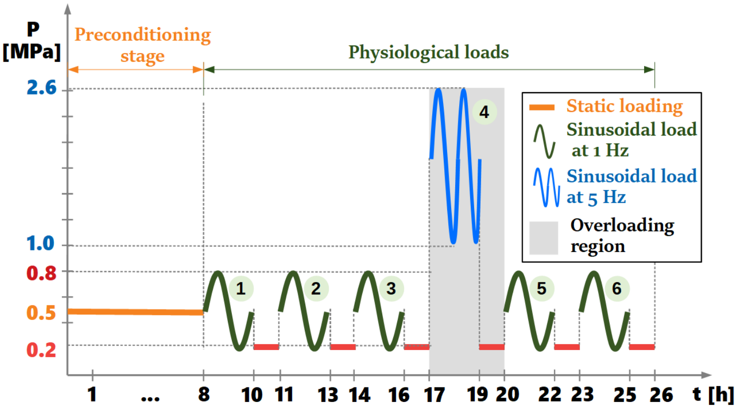


Fig. 1: Overloading regime that includes the loading/recovery periods of the test.

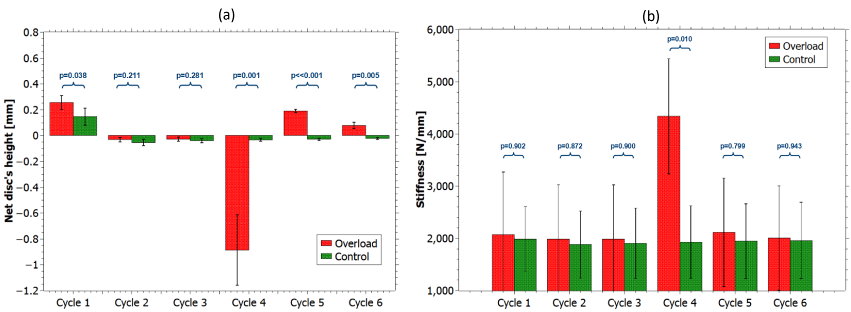


Fig. 2: Significance tests of the (a) Net disc's height and (b) axial stiffness.

Collagen Fiber Bundles Disintegration during Pull-out from the Endplate

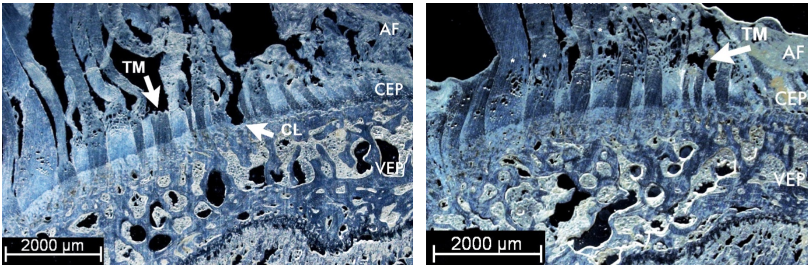
Wojtków M and Pezowicz C

Dept. of Biomedical Engineering, Mechatronics and Theory of Mechanisms, University of Science and Technology, Wroclaw, Poland

Rigid vertebra and flexible disc, functionally integrate with each other in the end-plate (EP) structure. In order to enable of junction these mechanically and structurally different tissues, each endplate consists of the cartilaginous endplate (CEP) and the vertebral endplate (VEP). Two layers of cartilage can be distinguished in EP- uncalcified and calcified, which is separated by the tidemark- calcification line (TM). Annular fiber bundles pass across TM, splits into sub-bundles and penetrates up to the cement line (CL)- boundary between CEP and VEP. Annuli-endplate junction ensures strong connection, used to transfer and distribute loads in spine. The aim of study was to analyze annuli-endplate anchorage, structurally and biomechanically, in physiological and degenerative state.

20 pigs lumbar motion segments (11 physiological, 9 degenerative), divided into samples including anterior and posterior region of tissues were tested. Degenerative changes were simulated by the endplate decalcification before tests. The multilayer systems of annuli were subjected to tensile using testing machine at a rate of 0.3mm/s until the rupture of annuli-endplate connection. In order to perform microstructural analysis tissues were fixed, decalcified and cut into 30µm thick slices (sagittal plane).

In the physiological state, TM failure was a dominant type of failure, what indicates that annuli-endplate anchorage is stronger than collagen fibers strength, therefore they are destroyed. High stresses generated in single sub-bundles at the CEP line, causes fiber bundles disintegration and failure of the whole bundle as a consequence. Moreover, TM and CL failure may occur simultaneously (Fig. 1). In this case, TM failure was observed at the outer region of annuli and pullout of fibers from CL at the inner annuli layers. In the degenerative state of endplate, CEP is damaged and doesn't work as a cement anymore, resulting in decreasing of the anchorage strength (failure force decline approximately by 30%) and pulling sub-bundles out from CL.



a b

Fig. 1: Microscopic images illustrating failure of the annuli-endplate anchorage: a) occurring at the tidemark and the cement line; b) disintegration of collagen fiber bundles indicated by * - shown as perforation of bundles (note: TM- tidemark, CL- cement line, AF- annuli fibrosus, CEP- cartilage endplate, VEP- vertebral endplate).

Voltage-Gated Ion Channels in Intervertebral Disc Mechanotransduction

Poillot P[a], O'Donnell J[a], Le Maitre C[b], Huyghe JM[a,c]

[a]*Bernal Institute, University of Limerick, Limerick, Ireland*
[b]*Biomedical Research Centre, Sheffield Hallam University, Sheffield, UK*
[c]*Dept. of Mechanical Engineering, Eindhoven University of Technology, Eindhoven, The Netherlands*

The Intervertebral disc is constantly subjected to forces generated by daily movement, primarily hydrostatic pressure in the nucleus and tension in the annulus. But disc degeneration can disrupt normal biomechanics, generating uneven and complex loading patterns. Evidence suggests that these forces are converted into voltages through mechanisms such as piezoelectricity, streaming potentials and electrostatic phenomena. This implicates voltage-gated ion channels in the biological remodelling response of the disc to loading, with potentially altered roles in degenerated tissue. While the mechanotransducive role of these channels has been extensively studied in chondrocytes, they have not been investigated in the disc. The focus of this study is to identify and determine the role of voltage-gated ion channels in healthy and degenerate disc mechanotransduction.

Piezoelectricity of annulus and nucleus will be investigated using a piezometer. Voltage-gated ion channels will be identified and localised by immunofluorescence on bovine and human tissue. The cell response to voltage will be examined by depolarization with 20mV and the response measured through immunofluorescence and RT-qPCR. Healthy and degenerate cells will be subjected to fluid-induced shear stress in the presence of specific channel inhibitors to elucidate their role in mechanotransudction.

Preliminary piezoelectricity testing indicates that annulus tissue exhibits the direct piezoelectric effect, while the nucleus does not. This may indicate that certain voltage-gated ion channels will have greater expression levels in the anulus during immunofluorescence.

As collagen fibres are aligned in the annulus, but disorganised in the nucleus, the piezoelectricity results align with the theory for structured biological systems. Other mechanisms, such as streaming potentials, may generate greater voltages in the nucleus. The localisation of voltage-gated ion channels through immunofluorescence will inform this hypothesis. Ultimately, only blocking these specific channels under a force will demonstrate the importance of such channels to disc maintenance and degeneration.

Intervertebral Disc Degeneration from a Biomechanical Point of View: What do we need to fix?

Vergroesen PPA[a,b,c], Rustenburg CME[b,c], Smit TH[b,c] , Emanuel KS[b,c]

[a]*University Medical Center Utrecht, Utrecht, The Netherlands*
[b]*Amsterdam Movement Sciences, Amsterdam, The Netherlands*
[c]*Amsterdam University Medical Center, Amsterdam, The Netherlands*

Since its early beginnings in the 1960s biomechanical research in the spine has matured into a multifaceted research field including a multitude of different disciplines. Although a body of research provided us with deeper insights into 'healthy' and 'degenerative' spine biomechanics, this has not yet led to adequate therapies. This work reviews current concepts on healthy and degenerative intervertebral disc biomechanics and provides a perspective on future directions for biomechanical research in the spine.

We searched Pubmed, Embase and Google Scholar for intervertebral disc, biomechanics, degeneration, regeneration and therapies. Inclusion criteria were works covering investigations into normal —'healthy'— or 'degenerative' biomechanics and therapeutic interventions relating or resulting in biomechanical stimuli. Exclusion criteria were cellular therapies, and injectables that did not affect biomechanical properties of the intervertebral disc.

From the combined works, we synthesized a glossary and a contemporary overview of current possible avenues of research. It was necessary to determine a glossary first as numerous often used terms lack a concise definition, which hampers understanding, especially across disciplines. Using the works included, we conclude that although a quick-fix is elusive, certain promising avenues of research are open. Especially promising are biomechanical stimulation within boundary's, restoration of a discs' fixed charge density, and restoration of the discs' permeability to water.

Unfortunately, little direct evidence exists on how to fix the intervertebral disc from a biomechanical point of view. Still, some promising directions are gleaned from current works, these include a balance between loading and unloading, restoration of the intervertebral discs' water binding capacity, and restoration of the permeability of the annulus and nucleus. However, when considering the degeneration of the spine as a vicious circle including biomechanics, cells and extracellular matrix, we must also intervene in the domains of cells and matrix to durably reverse degeneration.

Spine Postures, Physical Exposure, and Back Pain: a Systematic Review of Systematic Reviews

Swain Ch[a], Pan F[b], Owen P[c], Schmidt H[b], Belavy D[c]

[a]*School of Behavioural and Health Sciences, Australian Catholic University, Melbourne, Australia*

[b]*Julius Wolff Institute, Charité – Universitätsmedizin Berlin, Berlin, Germany*

[c]*Deakin University, Institute for Physical Activity and Nutrition, School of Exercise and Nutrition Sciences, Geelong, Victoria, Australia*

Low back pain (LBP) is the leading cause of disability worldwide. Mechanical and occupational factors, such as spine postures and specific physical exposures (e.g. prolonged sitting, bending, twisting, and lifting demands) have been strongly but variably linked to LBP, with previous reviews producing contrasting outcomes. Thus, despite many studies and reviews available, the evidence regarding the mechanical contributions to LBP is still inconclusive.

This umbrella review will examine 1) what relationship, if any, is evident between specific spinal postures or specific physical activities and back pain; 2) what is the quality of existing systematic reviews in this area; and 3) to what extent do existing reviews of mechanical factors and LBP demonstrate causality.

Five electronic databases and reference lists of relevant articles were searched from January 1990 to June 2018. Systematic reviews and Meta-Analyses on spine posture or physical exposure and back pain symptoms (self-report) or outcomes (e.g. work absence, medical consultation) were included. The AMSTAR and the Bradford Hill Criteria will be utilised by two independent reviewers to critically appraise the quality of included systematic reviews and to determine the extent to which these reviews demonstrate causality. This review has been prospectively registered on PROSPERO (CRD42018110739).

Of 6050 articles identified by the search, forty-three articles met the criteria to be included in the analysis. Although a variety of mechanical factors were positively associated with back pain, the quality of reviews was varied and several factors impact interpretation. This umbrella review will provide a comprehensive overview of the associations between mechanical factors and LBP.

Interactions between Genetics and Loading in Development of Disc Degeneration and Low Back Pain – a Review

Niedlinger-Wilke C[a], Kletsas D[b], Urban J[b]

[a]*University of Ulm, Institute of Orthopaedics and Biomechanics, Ulm, Germany*
[b]*National Centre of Scientific Research Demokritos, Athens, Greece*
[c]*DPAG, University of Oxford, Oxford, UK*

The large proteoglycan aggrecan, is a major component of the intervertebral disc and is made and maintained by disc cells. It is important for the disc's loading-bearing behaviour as, through imparting high osmolarity and low hydraulic permeability to the disc, it maintains disc turgor under varying external loads. Loss of aggrecan and hence swelling pressure occurs very early in disc degeneration, leading to loss of disc height and changes in stiffness. Loss also inhibits aggrecan biosynthesis, but induces disc cells to produce degradative proteases which destroy disc tissue, and also to produce inflammatory and pain-producing factors. Thus, as well as adversely affecting load-bearing, loss of aggrecan allows penetration of inappropriate macromolecules and invasion of blood vessels and nerves into the disc.

The relative roles of genetics and environmental factors in determining aggrecan loss and back pain are matters of debate and are reviewed here.

Degenerative changes in the disc appear load-induced in some cases as they are seen in sports where the spine is highly loaded (e.g. weight-lifting), the obese, and can be produced experimentally in vitro. However, independent of the magnitude of occupational spinal loading, twin studies find that the heritability of lumbar disc degeneration is very high, though the genes responsible have not been fully identified. Moreover the relationship between disc degeneration and pain is unclear. Imaging studies find many people with degenerate or herniated discs are asymptomatic. Is it that pain-producing degenerative features have not been identified on MRI, or is it that genetic differences to similar stimuli, affect responses to pain

Whether, and if so, how, genetics interacts with load-induced disc degeneration and herniations, and with production of and responses to inflammatory and painful stimuli, needs to be investigated and understood for development of preventative measures and design of rational back-pain treatments.

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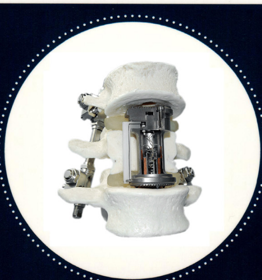

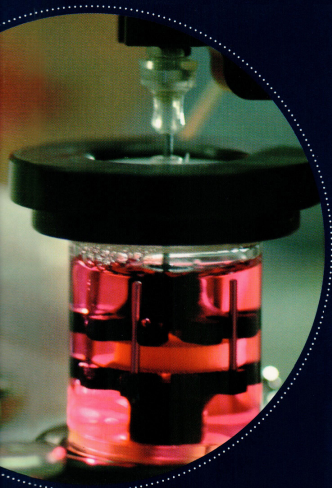

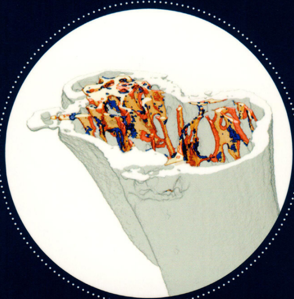

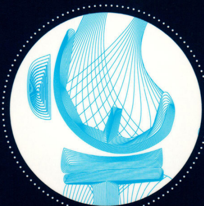

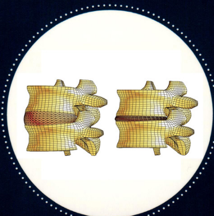

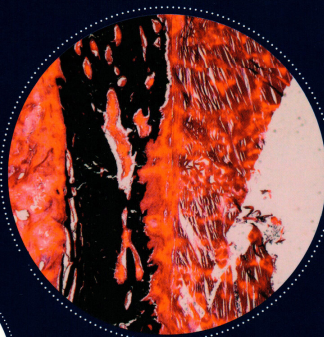

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Review of Load-sharing in Intact, Transected, Degenerate and Surgically Altered Passive Human Lumbar Spines

Ghezelbash F[a], Schmidt H[b], El-Rich M[c], Shirazi-Adl A[a]

[a]*Polytechnique Montréal, Montréal, Canada*
[b]*Julius Wolff Institute, Charité - Universitätsmedizin Berlin, Berlin, Germany*
[c]*Dept. of Mechanical Engineering, Khalifa University, Abu Dhabi, UAE*

Human lumbar motion segment is composed of various components (e.g., intervertebral disc, facet joints, ligaments) with distinct contributions to its mechanical response. By employing experimental/computational approaches, many studies have investigated the relative role of each component as well as effects of various factors such as boundary-initial conditions, load magnitude-combination-direction, load temporal regime, preload, posture, degeneration and surgical interventions (e.g., disc arthroplasty, posterior stabilization) on load sharing. This paper reviews and discusses the relevant findings of in vivo and in vitro observations as well as computational studies on load-sharing in healthy, aged, degenerated, damaged and surgically altered human lumbar motion segments.

We performed a specific search in PubMed using three sets of concepts ("lumbar spine", "load-sharing" and "motion segment components") and selected relevant papers as primary sources. References within primary sources were examined as secondary sources. Since not all studies have used the "load-sharing" term (or its equivalents), an additional generic search considering "lumbar spine", "motion segment components" and "biomechanics" in PubMed was performed. After two stages of evaluation, qualified papers were taken into account for a critical review.

In the initial selection process, 80 primary sources (out of 471) as well as 116 secondary sources were selected. 253 (out of 2133) studies were additionally included from generic search results. In the final stage of evaluation, all foregoing 449 papers were re-examined and nearly 250 studies were found qualified. Lumbar spine functional units (vertebra-disc-vertebra with no posterior elements) as well as single and multi-motion segments were considered.

In brief, the biomechanical role of each component was found to substantially alter with boundary conditions, geometry, load magnitude-combination, preload compression, disc hydration, posture, bone quality and time (creep and repetition). Transection order affects findings and conclusions not only in force-control protocols but also in fully displacement-control loading regimes. Disc degeneration, endplate fracture, nucleotomy and surgical interventions significantly alter load transmission in the lumbar spine. Flexibility of posterior elements and geometry/placement of interbody cage are influential variables affecting flexibility and load-sharing of the spine in fusion surgery.

Relationship between Intervertebral Disc and Facet Joint Degeneration: a Probabilistic Finite Element Model Study

Bashkuev M, Reitmaier S, Schmidt H

Julius Wolff Institute, Charité – Universitätsmedizin Berlin, Berlin, Germany

Both intervertebral disc and facet joint degeneration are often believed to be common causes of chronic low back pain. While the biology and the genetics are considered the most relevant factors in the onset of degeneration, the role of mechanics in progression of degeneration is evident [1, 2]. Degenerative changes in one of the structures are believed to induce degeneration in the other. However, despite decades of research, there is no consensus on the mechanical interplay between the structures.

Based on a parametric finite element model of a human L4-5 spinal motion segment, one thousand individual segments were probabilistically generated covering all grades of degeneration in both structures (Fig. 1). The segments were subjected to combined compression and flexion/extension loads. Correlation matrices were created to identify the effect of individual degeneration parameters (e.g., disc height, facet gap width) on the mechanical stresses in the opposite structure.

In non-degenerated group, strong and moderate negative correlation was found between the strain of the capsular ligament and the disc height loss and the nucleus compressibility, respectively (Fig. 2). In mild degeneration, the correlation between the disc height loss and the capsular strain remained strong, while a moderate correlation emerged between the facet gap width and the force on the intervertebral disc. With increasing degeneration, no correlations were found between the individual intervertebral disc morphologies and the facet joint loads. The disc load, in turn, showed strong correlation with the facet gap and moderate with the cartilage stiffness.

The results suggest that early stages of disc degeneration have the largest effect on the facet joint loading. With progression of degeneration, this effect diminished, while the appearances of facet joint degeneration gain importance and likely support the disc degeneration.

[1] Iatridis et al, Spine J. 2013. Role of biomechanics in intervertebral disc degeneration and regenerative therapies: What needs repairing in the disc and what are promising biomaterials for its repair?

[2] Iorio et al, Asian Spine J. 2016. Biomechanics of Degenerative Spinal Disorders.

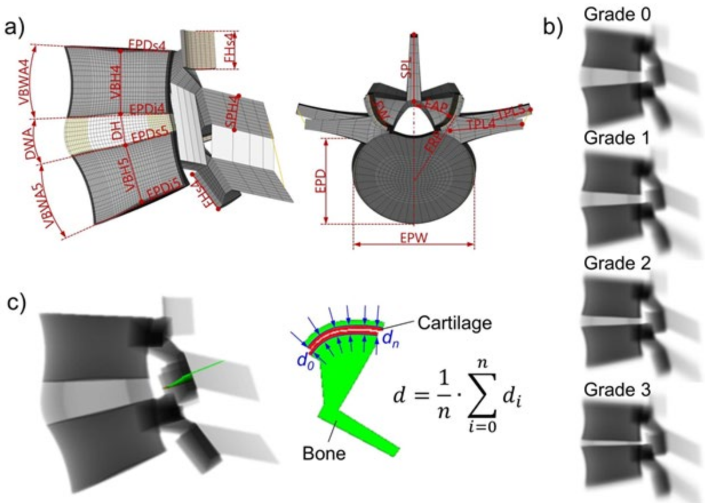


Fig. 1: a) Parametric finite element model of a human L4-L5 motion segment. b) Simulated “X-rays” (inverted) of the models representing different stages of degeneration (averaged images of individual models for each group). c) The facet joint gap is determined as the mean distance between the opposing bony surfaces of the facets in the cutting plane through the middle of the facet joint.

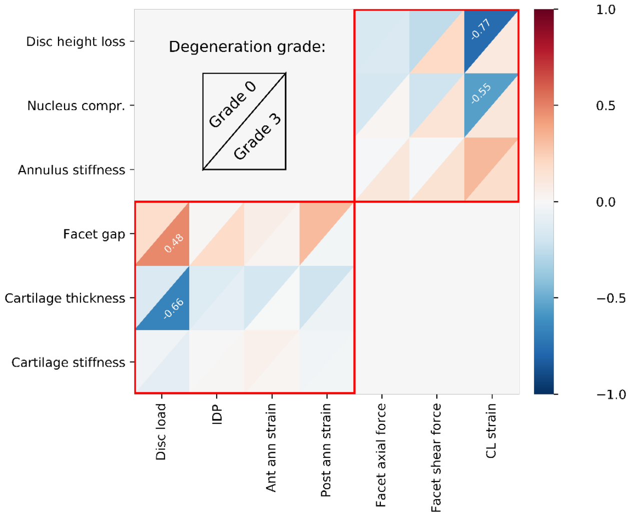


Fig. 2: Heat map of the correlation matrix of the degeneration parameters and evaluated values for the non-degenerated / severely degenerated group. Weak and no correlation values are not annotated.

Lumbar Spinal Ligament Characteristics extracted from Stepwise Reduction Experiments allow for Precise Modeling

Damm N[a], Rockenfeller R[b], Gruber K[a]

[a]*MTI Mittelrhein, Institute for Medical Engineering and Information Processing, University of Koblenz-Landau, Campus Koblenz, Koblenz, Germany*

[b]*Mathematical Institute, University of Koblenz-Landau, Campus Koblenz, Koblenz, Germany*

Lumbar ligaments play a key role in stabilizing the spine, particularly assisting muscles at wide-range movements. Hence, valid ligament force-strain data are required to generate physiological model predictions. These data have been obtained by experiments on single ligaments or functional units throughout literature. However, contrary to detailed spine geometries, gained for instance from CT data, ligament characteristics are often inattentively transferred to multi-body system (MBS) or finite element models.

We use an elaborated MBS model of the lumbar spine to demonstrate how individualized ligament characteristics can be obtained by reversely reenacting stepwise reduction experiments, where the range of motion (ROM) was measured. We additionally validated the extracted characteristics with physiological experiments on intradiscal pressure (IDP).

Our results on a total of in each case 160 ROM and 49 IDP simulations indicated a clear superiority of our procedure (seven and eight outliers) towards the incorporation of classical literature data (on average 71 and 31 outliers).

Reasons for the observed variance in ligament characteristics are that (i) the individual ligaments are more or less strongly grown together in cadaver experiments, (ii) insertion points of ligaments are often planar, but must be reduced to a single point in the model and (iii) the beginning and the end of some ligaments is hard to define, since some fibers of the lig. longitudinale anterius, lig. longitudinale posterius and lig. supraspinale pass over several vertebral bodies. It should finally be noted that data sets are often not complete.

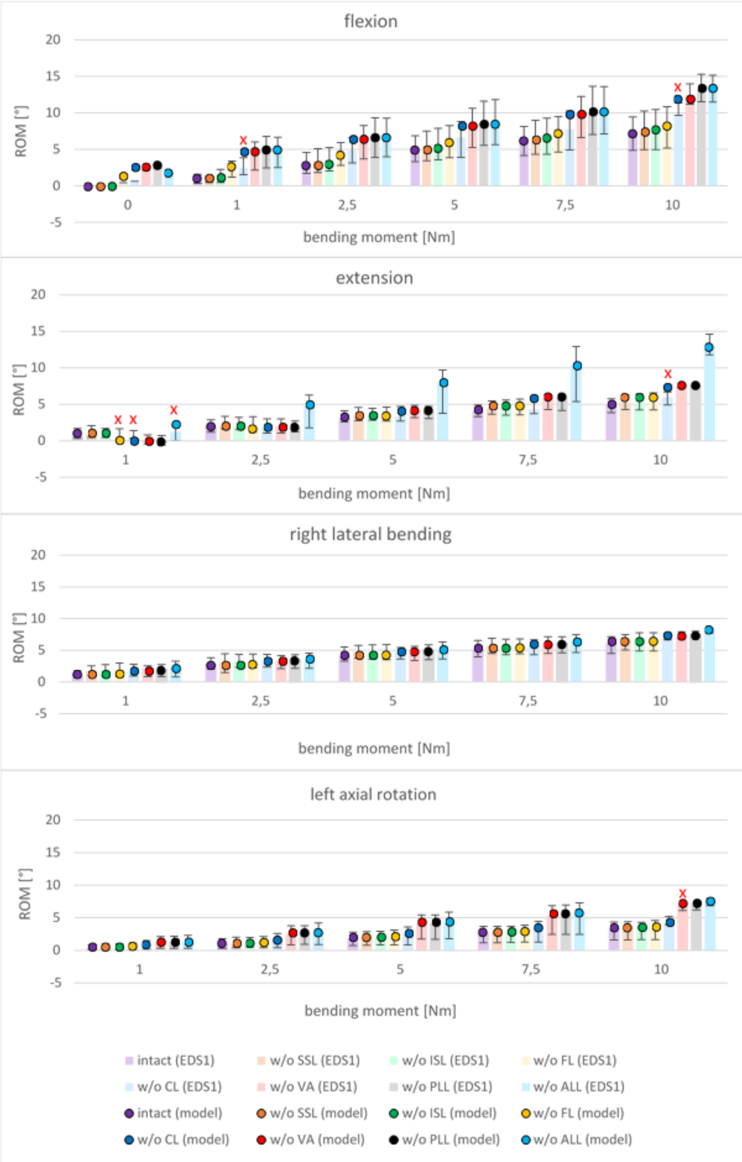


Fig. 1: Bar plots visualizing the experimentally determined ROM of the stepwise reduction experiment by Heuer et al. (2007) at different torques along all three anatomical planes. Colored bars represent mean ROM values for different reduction steps. Black error bars indicate the ranges (minimum to maximum). The correspondingly colored circles show the simulation results of our L4-L5 model using the extracted IVD- and ligament characteristics. Red crosses show the seven outliers. Note that there were no ranges given in the neutral position (0 Nm).

Detailed Full-filed Analysis of the Ventral Lumbar Spine: Insights on the Biomechanical Role of the Anterior Longitudinal Ligament

Palanca M[a], Ruspi ML[a], Cristofolini L[a], Liebsch C[b], Villa T[c,d], Brayda-Bruno M[d], Galbusera F[d], Wilke HJ[b], La Barbera L[c,e]

[a]*Dept. of Industrial Engineering, School of Engineering and Architecture, Alma Mater Studiorum – Università di Bologna, Bologna, Italy*
[b]*Institute of Orthopaedic Research and Biomechanics, Trauma Research Center Ulm (ZTF), University Hospital Ulm, Ulm, Germany*
[c]*Laboratory of Biological Structure Mechanics, Department of Chemistry, Materials and Chemical Engineering "G. Natta", Politecnico di Milano, Milan, Italy*
[d]*IRCCS Istituto Ortopedico Galeazzi, Milan, Italy*
[e]*École Polytechnique de Montréal, CHU Sainte-Justine, Montréal, Canada*

The anterior longitudinal ligament (ALL) has a fundamental role in constraining the motions in the sagittal plane. Its specific contribution has been studied only as part of a whole functional spinal unit and with simple tensile tests on isolated specimens. The aim of this study is, therefore, to investigate in depth the biomechanical role of the ALL and how it is strained both in front of the lumbar vertebrae and of the intervertebral disc (IVD) under pseudo-physiological conditions as part of a multi-vertebra spine segment.

Five thoracolumbar spine specimens were subjected to *in vitro* tests using a state-of-the-art spine tester. Before testing, a white-on-black speckle pattern was prepared on the specimens (Fig. 1). A commercial digital image correlation (DIC) system was deployed to measure the full-field displacements and strains distribution on the ventral part during flexibility tests. Unconstrained pure-moments (±7.5 Nm) were sequentially applied in flexion-extension, lateral bending and axial torsion. Preliminary CT scans allowed to identify specific bony features, eventually correlated with any local detail in the strain maps.

In extension, the ALL bundles stretched axially (Fig. 2,§,+). In flexion, the ALL bundles were mainly stretched circumferentially (Fig. 2,#) in front of the intervertebral disc (IVD) due to disc pressurization/bulging. In lateral bending similar effects are observed at the concave and convex sides undergoing compression (Fig. 2,#) and tension (Fig. 2,§,+), respectively. In torsion, the ALL bundles were stretched (Fig. 2,*) at roughly 45° with respect to the spine axis.

The biomechanical behavior of the ALL has been comprehensively evaluated for different loading conditions. The measured strain maps showed highly inhomogeneous and typically non-symmetric distributions during the different loading conditions. The ALL was generally more strained in front of the IVD rather than in front of the vertebrae, with local strain intensification imputable to local bony defects (e.g. osteophytes).

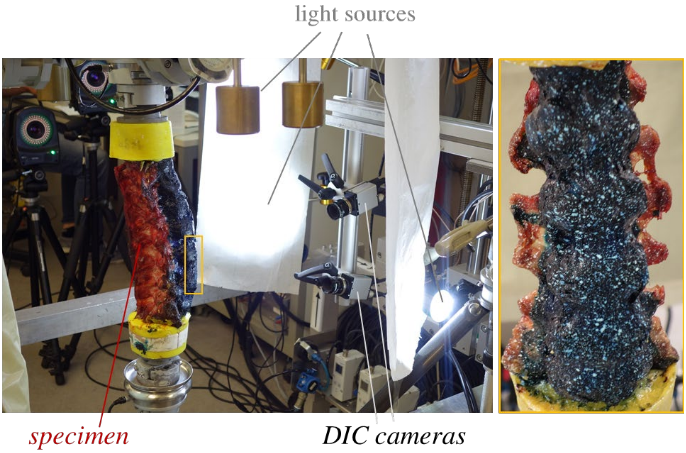


Fig. 1: Details of the test set-up for 3D-DIC analysis of the ventral spine of a specimen prepared with a white-on-black speckle patter. Left: The specimen was mounted on the spine tester: two DIC cameras were positioned in front of the specimen with high-intensity light sources to improve image quality.
Right: detail of the white pattern allowing to track the displacements and deformations of the hard and soft tissues of the spine; the pattern was obtained spraying a water-based white paint with an optimized airgun on a methylene-blue background.

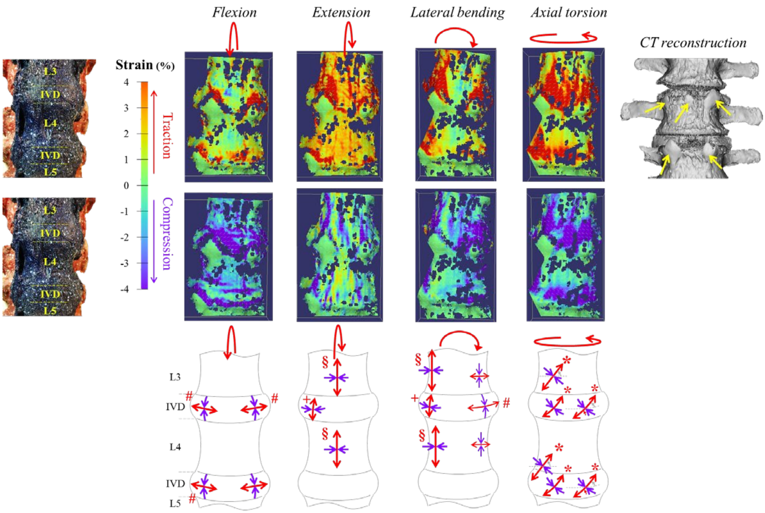


Fig. 2: Examples of maximum (top) and minimum (mid) principal strains maps measured with DIC on the ventral spine of a representative specimen in flexion, extension, left lateral bending and left axial torsion. Schematic drawings (bottom) to visualize the alignment of the principal strains in the different regions reported in the strain maps. A photo of the ventral spine helps identifying each region of interest (left), while the CT reconstruction (top right) clearly shows the presence of osteophytes where high strain intensification occurred.

Effects of Lumbar Lordosis on Mechanical Response of Post-operative Lumbar Spine - Personalized Parametric Finite Element Simulations

Nikkhoo M[a], Khoz Z[a], Cheng CH[b,c], Niu CC[c,d], El-Rich M[e], Almansouri S[f], Hebela N[f], Khalaf K[g]

[a]*Dept. of Biomedical Engineering, Science and Research Branch, Islamic Azad University, Tehran, Iran*
[b]*School of Physical Therapy and Graduate Institute of Rehabilitation Science, College of Medicine, Chang Gung University, Taoyuan, Taiwan*
[c]*Bone and Joint Research Center, Chang Gung Memorial Hospital, Linkou, Taiwan*
[d]*Dept. of Orthopaedic Surgery, Chang Gung Memorial Hospital, Linkou, Taiwan*
[e]*Dept. of Mechanical Engineering, Khalifa University, Abu Dhabi, UAE*
[f]*Orthopaedic Spine Surgery, Neurological Institute, Cleveland Clinic Abu Dhabi, Abu Dhabi, UAE*
[g]*Dept. of Biomedical Engineering, Khalifa University, Abu Dhabi, UAE*

Lumbar interbody fusion includes various techniques that primarily aim to achieve fusion across the anterior column of the intervertebral disc. Clinically, it's also important to correct the abnormal lordosis angle (LA) and restore sagittal balance [1]. Estimation of the effect of the LA on spinal biomechanics after fusion is critical for proper surgical planning. This study aims to (1) develop a validated personalized finite element (FE) model that automatically updates spinal geometry for different patients; and (2) use the model to study the influence of the LA on spinal biomechanics post fusion.

Using an X-Ray image-based user-defined code, the geometry of the lumbar spine (L1-S1) was automatically updated by independent parameters. Fifteen subject-specific nonlinear osteoligamentous FE models were developed based on pre-operative images of patients (Fig. 1). Post-operative FE models of the same patients were also created (Fig. 2). A parametric study of the effect of the LA in fusion was investigated for cases with no change in the LA, increased LA (+6° and +12°) and decreased LA (-3° and -6°), producing a total of 75 fusion models. The results were compared using one-way ANOVA.

The intersegmental ranges of motion (ROMs), intradiscal pressure (IDP), and facet joint forces (FJF) for the pre-operative models were consistent with literature [2-4]. The average ROM, IDP, adjacent stress levels, and interbody cage stress were significantly higher with decreased LA during flexion, extension, and lateral bending, but no significant changes were detected during axial rotation. There were no significant changes in the FJF.

The novel personalized FE models developed in this study provide a simple, and cost-effective analysis of the biomechanical changes associated with lumbar spinal fusion. This study demonstrates that the LA alters both the intersegmental motion and load sharing in fusion, which may influence the initiation and rate of adjacent level degeneration.

[1] Blizzard et al., Int. J. Spine Surg., 2016 (29)
[2] Panjabi et al., J. Bone Joint Surgury, 1994(76)
[3] Brinckmann et al., Spine, 1991 (16)
[4] Wilson et al., J. Biomechanics, 2006 (39)

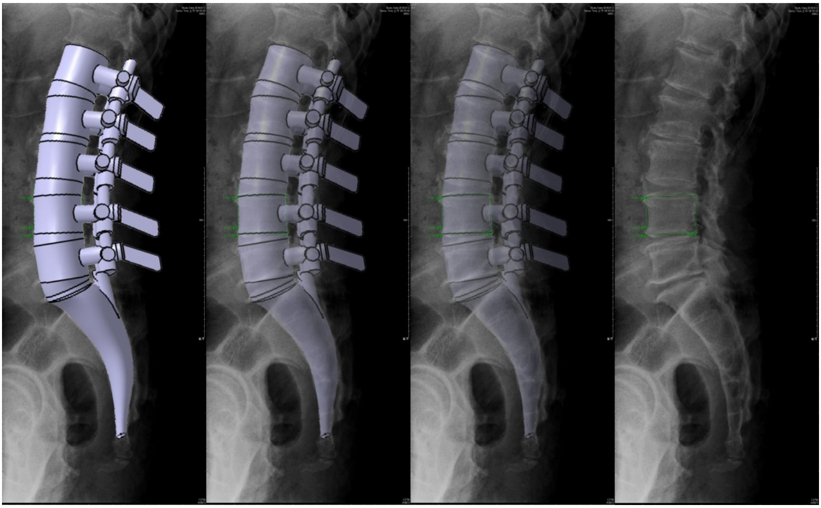


Fig. 1: The geometrical model of pre-operation for patient No.1.

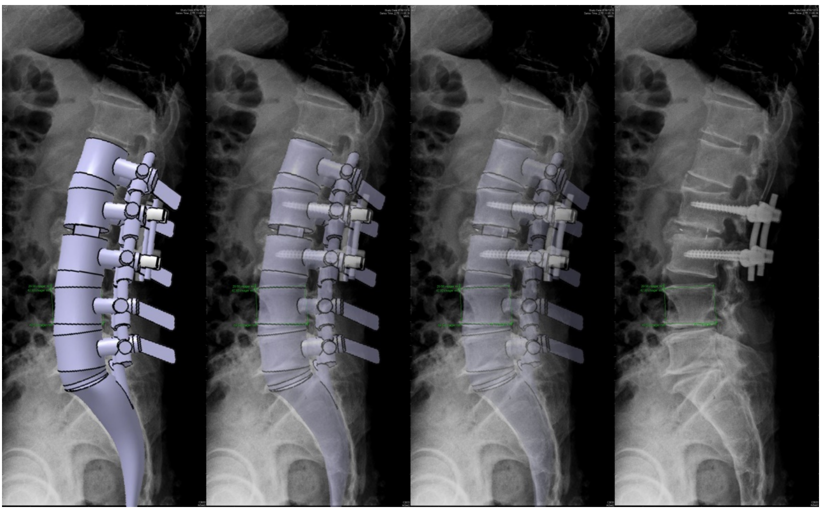


Fig. 2: The geometrical model of post-operation for patient No.1 (Fusion at L2-L3 Level).

Biomechanical Properties in Motion of Lumbar Spines with Degenerative Scoliosis

Rustenburg CME[a], van der Veen A[a], Holewijn RM[b], Faraj SSA[c], Kingma I[d], Smit TH[e], Emanuel KS[a]

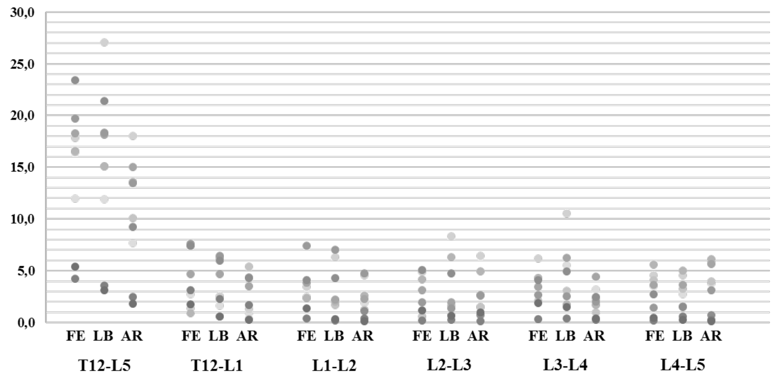
[a]*Dept. of Orthopaedic Surgery, Amsterdam UMC, Amsterdam Movement Sciences, Amsterdam, The Netherlands*
[b]*Dept. of Radiology, OLVG Amsterdam, Amsterdam, The Netherlands*
[c]*Dept. of Orthopaedic Surgery, Radboud UMC, Nijmegen, The Netherlands*
[d]*Dept. of Human Movement Sciences, Vrije Universiteit Amsterdam, Amsterdam, The Netherlands*
[e]*Dept. of Medical Biology, Amsterdam UMC, Amsterdam Movement Sciences, Amsterdam, The Netherlands*

Degenerative lumbar scoliosis results in malalignment of the spine and presumably alters spinal biomechanics. However, there is a lack of quantitative reference measurements of spines with degenerative scoliosis, which are needed for modeling and evaluation of treatments. Therefore, we aimed to assess the biomechanical properties in motion of spines with degenerative scoliosis, and to relate these properties to intervertebral disc degeneration (DD) and Cobb angle. Secondly, we compared these results to previous measurements of non-scoliotic spines.

Ten lumbar cadaveric spines (Th12-L5, age 82 ±11) with a Cobb angle ≥10° and apex on L3 were acquired. Three loading cycles from -4 to 4 Nm were applied per direction, in flexion and extension (FE), lateral bending (LB), and axial rotation (AR). The ROM and neutral zone (NZ) stiffness were calculated for each motion segment in the direction of loading. Additionally, ROM was calculated in coupled directions.

For T12-L5, there was a ROM (degrees ±SD) of 14.88 (±6.45) in FE, 14.85 (±7.80) in LB, and 10.15 (±5.52) in AR, and the median (Nm/degree (Q1;Q3)) NZs was 0.24 (0.19;0.35) in FE, 0.25 (0.22;0.42) in LB, and 0.49 (0.33;0.99) in AR. Largest coupled motions were obtained in LB during FE-loading on L2-L3 (median 125.24%, Q1;Q3 86.86;176,35), expressed as a percentage of rotation in the loaded direction. No differences between spinal levels and no correlations with Cobb-angle were observed. DD correlated to lower ROM on L2-L3 in FE ($R=-0.762$, $p=0.027$) and increased NZ stiffness on L3-L4 in LB ($R=0.669$, $p=0.049$). Compared to non-scoliotic reference spines, significantly smaller ROM in FE ($p=0.030$) was found.

This study describes the natural envelope of motion in lumbar spines with degenerative lumbar scoliosis. Compared to non-scoliotic spines, spines with degenerative scoliosis tended to be stiffer, with smaller ROM in FE. DD only affected the ROM and NZ stiffness around the apex.



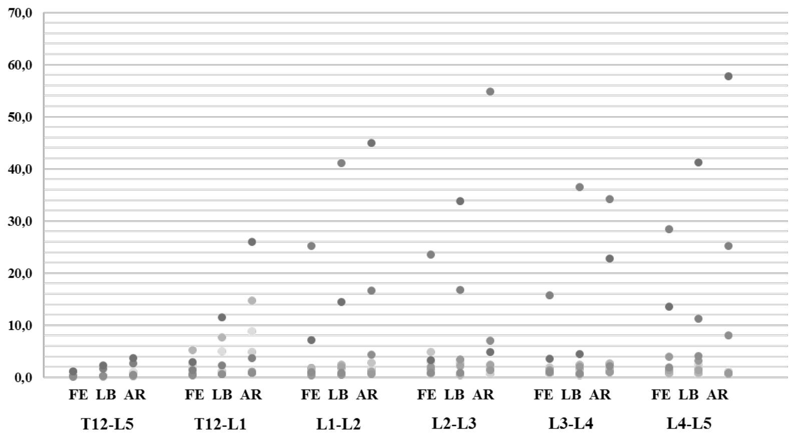


Fig. 1: The ROM (in degrees; upper diagram) and NZ stiffness (in Nm/degree; lower diagram) for each spine, for the whole spine and per segment, and per direction.

Review Article on Spine Kinematics of Quadrupeds and Bipeds

Reitmaier S, Hu Z, Pan F, Schmidt H

Julius Wolff Institute, Charité - Universitätsmedizin Berlin, Berlin, Germany

Due to their long history of use and despite the absence of sufficient proof, large animal 'models', mainly sheep, pigs and goats, are often assumed to be valid for preclinical spinal applications. The major arguments to justify their use are currently founded merely in *in vitro* studies demonstrating similarities in spinal morphometry and flexibility between animals and humans. This line of argumentation, however, is insufficient to vouch for the suitability of large animal models. Valid and reliable conclusions from animal studies for humans are only allowed if *in vivo* spinal mechanical attributes are comparable to humans. The present paper reviews existing literature on *in vivo* spinal kinematics of large quadrupeds and primates and critically discusses the comparability between these species and humans and thus their suitability for preclinical studies.

A literature review on the *in vivo* spinal kinematics of large quadrupeds, primates and humans was performed via electronic searching of relevant literature in suitable databases (PubMed, Embase, Web of Science, Google scholar). The search strategy was complemented by searching the references of qualified articles. The following inclusion criteria were applied: i) detailed specification of spinal range of motion/flexibility (cervical, thoracic and lumbar), ii) *in vivo* measurements, iii) common quadrupedal animal species and primates (i.e., exclusion of exotic species). For better inter-species comparison, kinematic data were divided into measurements during standard choreographies of standing subjects *vs.* moving subjects.

The extensive literature search is not yet completed. Up to now we included 75 articles. Most studies were found for horses (n=39) and humans (n=24). Additionally, kinematic studies on dogs (n=7) and primates (n=4; in both, bi- and quadrupedal gait) were included. Except one study for the standing sheep, no kinematic studies could be found investigating the spinal flexibility of the moving sheep, pig and goat. The collected kinematic data are very heterogeneous. Reasons for the high level of heterogeneity include anatomical differences in the number of thoracic and lumbar vertebrae between the different species, methodological differences, investigated spinal levels etc. Preliminary data on equine and human spinal ranges of motion reveal a velocity dependence with locomotion. Whereas the flexibility of the equine spine in flexion/extension, lateral bending and axial rotation markedly decreases from walk (≤2m/s) to trot (>2m/s), the flexibility of the human spine increases between comparable speeds of locomotion. Similar to humans, for primates walking bipedally, e.g., the Japanese macaque and the chimpanzee, a slight trend towards increased spinal flexibility with increasing speed of locomotion was observed. Due to the small number of animals, however, this is without statistical significance.

Overall, the characterization of the quadrupedal spine is far from complete. Except of the horse, kinematic *in vivo* data, especially of the various large animal models used for the preclinical spinal research are sparse or lacking. More *in vivo* studies are urgently needed to characterize the quadrupedal spine in order to prove or refute the significance of those animals in translational medicine.

Dynamic Interactions between Lumbar Intervertebral Motion Segments during Forward Bending

Breen Alexander[a] and Breen Alan[b]

[a]*Centre for Biomechanics Research, AECC University College, Bournemouth, UK*
[b]*Faculty of Science and Technology, Bournemouth University, Bournemouth, UK*

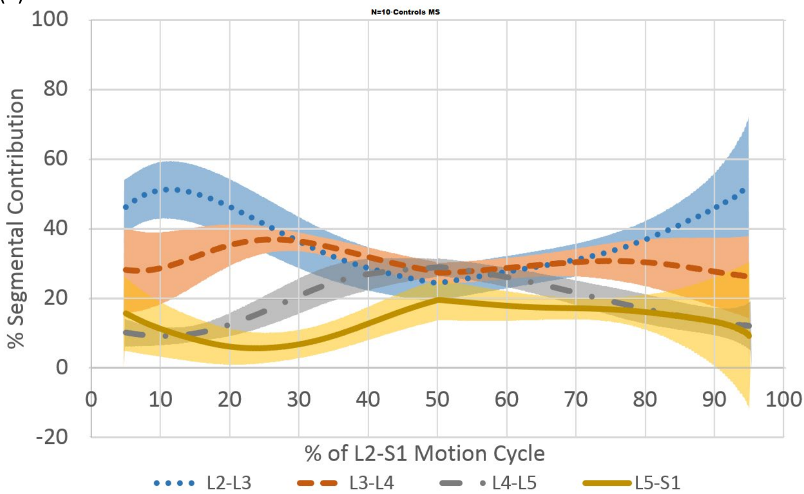
Surface marker studies have found greater flexion ranges in the upper than lower lumbar spine in patients with nonspecific low back pain (NSLBP) than controls, however, these do not describe the dynamic interactions between motion segments. Subsequent studies using quantitative fluoroscopy (QF) have found that intersegmental motion is more unequally shared in NSLBP than controls but did not attribute this to individual segments. The purpose of the present research was to describe motion sharing inequality (MSI) in terms of restraint and variability at individual segments (L2-S1) throughout bending in patients and controls to inform multi-segmental dynamic loading models in back pain.

One hundred and one pain free volunteers received QF during controlled lumbar flexion. Dynamic motion sharing of segments from L2-S1 and their MSI were calculated along with correlation coefficients between MSI and IV-RoM for each level. Ten controls were then matched to 10 patients with NSLBP for age and sex, and their MSIs and dynamic motion sharing patterns compared.

The study of controls (n=101) found the share of motion was highest at L2-3 and L3-4 and lowest at L5-S1 throughout the motion. This was exaggerated with higher MSIs. The second study (n=20), found that patients had non-significantly higher MSI's than controls, (p=0.17) and significantly higher proportional IV-RoMs at L2-3 and L3-4 than at L5-S1 (p<0.01). The proportional sharing of motion was also less variable throughout the sequences at L2-3 and L4-5 in patients (see Fig.).

Intervertebral motion sharing inequality is a normal feature during lumbar flexion and is characterised by increased motion at L2-3 and L3-4 and decreased motion at L5-S1. However, in patients with CNSLBP, this is more pronounced, and associated with less variation at some levels. These effects may result from changes in muscular contraction or in the mechanical properties of the disc.

(a)



(b)

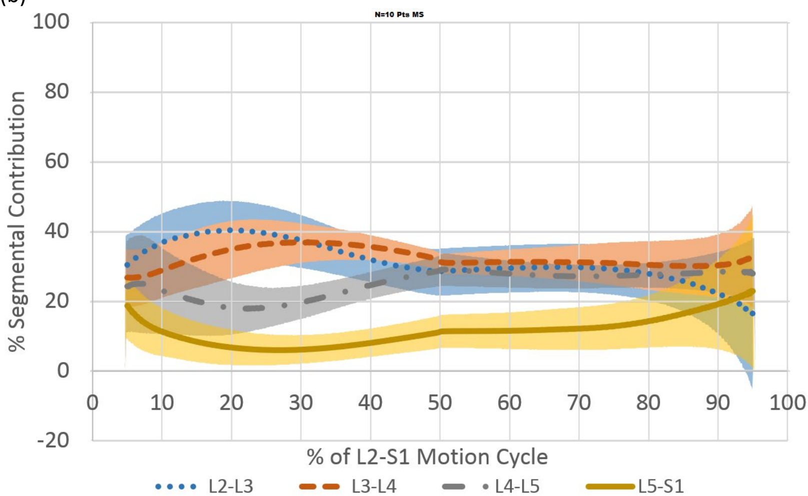


Fig.: Continuous intersegmental contributions (L2-S1) to forward bending motion in 10 healthy controls (a) and 10 patients with CNSLBP (b) (shading = 95% CI).

Sex-dependent Difference in Lumbo-Pelvic Coordination for Different Lifting Tasks

Pan F, Firouzabadi A, Zander T, Schmidt H

Julius Wolff Institute, Charité – Universitätsmedizin Berlin, Berlin, Germany

Manual material lifting is considered a risk factor for low back pain (LBP) [1]. During lifting, the sagittal motion is achieved through the lumbo-pelvic coordination, which is quantified by the ratio between lumbar and hip rotation (L/P ratio) [2]. Detailed knowledge about lumbo-pelvic coordination is a prerequisite for LBP analysis. Previous study demonstrated difference in LBP prevalence between sexes during occupational lifting activities [3]. However, the sex-dependent difference in L/P ratio has not yet been investigated.

An optoelectronic system (Vicon, Oxford, GB) was utilized to measure the lumbo-pelvic motion in 20 subjects (10m, 10f). Task A is lifting one weight from the ground in front of the body to three target heights with straight knees (A1: Level of abdomen; A2: Level of chest; A3: Level of head). Task B is lifting two equal weights from the ground at both sides of the body to three target angles with bended knees (B1: Arms close to the trunk; B2: Arms 45° abducted; B3: Arms 90° abducted). 10 kg (m & f) and 20 kg (m only) lifting were performed and three phases were investigated: Phase 1 – Upper body flexion to reach weights; Phase 2 – Lifting up weights; Phase 3 – Lowering down weights.

During both tasks, females normally displayed a smaller L/P ratio than males. In phase 2 and 3, the L/P ratio was greater than in phase 1. During task B, L/P ratio increased with an increasing lifting height. Different lifting weights displayed no difference in L/P ratio (Fig. 1). These results can partly explain the sex-dependent difference in LBP prevalence and can further provide indications for subject-specific recommendations for safer lifting activities.

[1] Parreira P, et al. Spine J 2018, 18:1715-1721.
[2] Pries E, et al. J Biomech 2015, 48:3080-3087.
[3] Freburger JK, et al. Arch Intern Med 2009, 169:251-258.

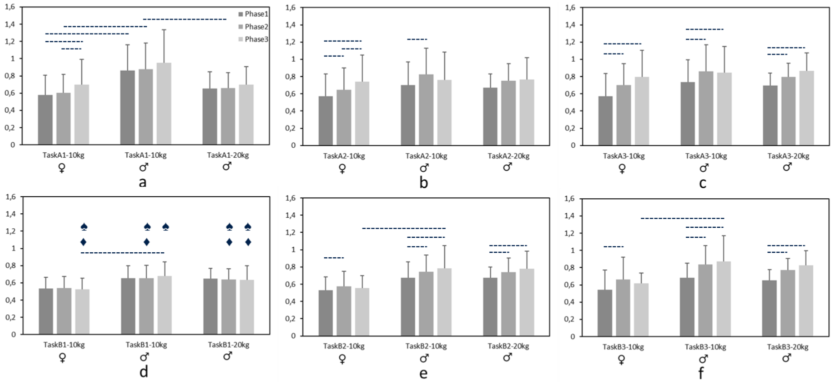


Fig. 1: L/P ratio (mean ± standard deviation) of task A – Lifting a weight from 10 cm in front of the body on the ground to three target heights: a. Level of abdomen, b. Level of chest, c. Level of head. L/P ratio of task B – Lifting weights from 10 cm at both sides of each foot on the ground to three target angles: d. Arms close to the trunk, e. Arms 45° abducted, f. Arms 90° abducted. Phase 1: Upper body flexion without weight; Phase 2: Lifting up weights; Phase 3: Lowering weights. ♀ Females, ♂ Males. Dash line $p<0.05$; ♠ – Comparison to height 3 in same sex and lifting weight group $p<0.05$; ♦ – Comparison to height 2 in same sex and lifting weight group $p<0.05$.

A Novel Model and Experimental Validation Demonstrate the Large Contribution of Passive Muscle to Spine Flexion Relaxation

Zwambag DP[a]* and Brown SHM[a]

[a]*Dept. of Human Health and Nutritional Sciences, University of Guelph, Guelph, Canada*
**Currently in Dept. of Kinesiology and Physical Education, Wilfrid Laurier University, Waterloo, Canada*

Flexion Relaxation (FR)—extensor muscle inactivation near full trunk flexion [1]—is believed to occur when passive tissues fully support the moment of the upper body. The relative contributions of disc, ligaments, and passive muscle to FR are unknown. This study was designed to estimate and compare the contributions of various structures supporting the spine during trunk flexion. It was hypothesized that the combined passive structures would resist the entire sagittal moment near full flexion. The validity of model predictions was further probed by manipulating the sagittal moment.

Ten healthy participants performed trunk flexion movements. A pulley-system was used to alter the sagittal moment; the L4/L5 sagittal moment was estimated using a rigid linked-segment, inverse dynamic analysis. Spine kinematics were supplied to an anatomically detailed computational model [2] to predict the lengths of 14 ligaments and 48 muscles crossing the L4/L5 disc. Disc and ligament moments were predicted using established models [3, 4]. Passive muscle moments were predicted using a custom model [5]. Briefly, human cadaveric sarcomere lengths [6] were scaled by muscle strain and used to predict passive muscle forces from the experimentally derived stress-sarcomere length relationship of rabbit multifidus fibre bundles [5]. The difference between the sagittal and combined passive tissue moments predicted the active muscle moment, which was compared to recorded muscle activity.

The model correctly predicted FR. At full flexion, the predicted active muscle moment was <2 Nm. At the instant of FR, disc, ligaments, and passive muscle supported 16%, 30%, and 44% of the sagittal moment, respectively. The model performed well when the sagittal moment was reduced, predicting less extensor and greater abdominal activity.

The model highlights that near full flexion, passive muscle moments are substantial and support the greatest proportion of the sagittal moment, although ligaments are stiffer and sensitive to small changes in spine flexion angle.

[1] Floyd WF, Silver PHS. 1951. The function of the erectors spinae in flexion of the trunk. Lancet 1; 133-134.
[2] Cholewicki J, McGill SM. 1996. Mechanical stability of the in vivo lumbar spine: implications for injury and chronic low back pain. Clin Biomech 11; 1-15.
[3] Adams MA, Dolan P. 1991. A technique for quantifying the bending moment acting on the lumbar spine in vivo. J Biomech 24; 117-126.
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lifts with varying degrees of trunk flexion. Spine 16; 1099-1107.
[5] Zwambag DP, Gsell KY, Brown SHM. 2019. Characterization of the passive mechanical properties of spine muscles across species. J Biomech (in review)
[6] Zwambag DP, Ricketts TA, Brown SHM. 2014. Sarcomere length organization as a design for cooperative function amongst all lumbar spine muscles. J Biomech 47; 3087-3093.

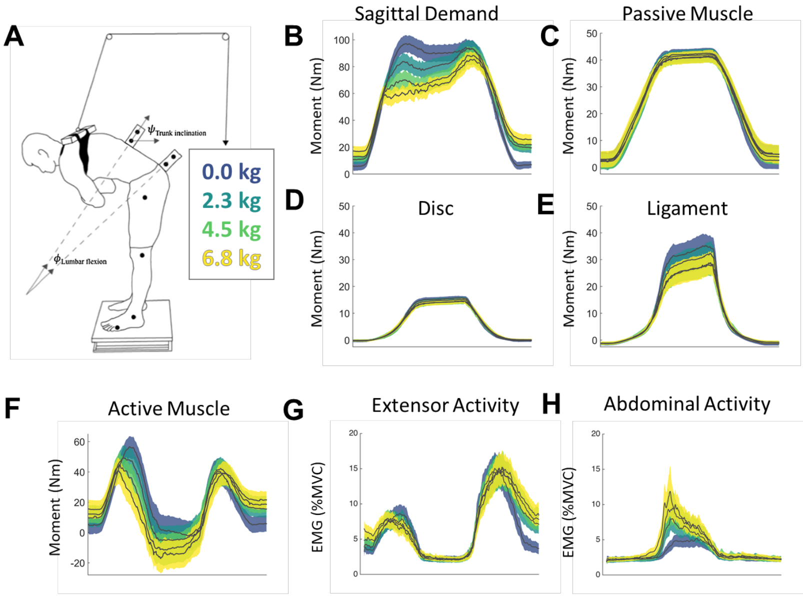

Fig. 1: (A) Schematic of experimental set-up showing the pulley system used to off-load mass from the torso. (B) Mean sagittal moment (shaded regions are 95% confidence in the mean estimate) for each loading condition throughout the movement. (C-E) Estimates of extensor moments generated by passive spine structures. (F) Active muscle moment predicted by the model as the difference between the sagittal moment and the combined passive structures. Positive or negative values indicate extensor or abdominal muscles are required, respectively. (G-H) Recorded EMG of extensor and abdominal muscle activity.

Calculating the Three-dimensional Vertebral Orientation from a Planar Radiograph: is it feasible?

Galbusera F[a], Niemeyer F[b], Wilke HJ[b], Bassani T[a], Sconfienza LM[c]

[a]*Laboratory of Biological Structures Mechanics, IRCCS Istituto Ortopedico Galeazzi, Milan, Italy*

[b]*Institute of Orthopaedic Research and Biomechanics, Trauma Research Center Ulm, Ulm University, Ulm, Germany*

[c]*Dept. of Diagnostic and Interventional Radiology, IRCCS Istituto Ortopedico Galeazzi, Milan, Italy*

In this study, we trained a deep neural network for the three-dimensional estimation of the direction of the three main anatomical axes (cranio-caudal, anteroposterior and laterolateral) of lumbar vertebrae from a single sagittal radiographic image taken from an approximately lateral direction with non-negligible deviations from a perfect alignment up to 30-40 degrees. To this aim, we exploited the three dimensional nature of computed tomography (CT) datasets, which can be used to accurately locate the position of anatomical landmarks and to create a high number of simulated radiographic projections with different orientations, for the creation of large training and validation datasets. A set of 21 CT stacks of patients were retrospectively collected. By using in-house software, the location of 5 landmark points was manually determined in the CT datasets for L2, L3 and L4, for a total of 63 vertebrae. For each vertebra, 200 simulated projections approximately aligned with sagittal plane but including random perturbations of the projection direction were built. The procedure resulted therefore in the generation of 12600 simulated radiographs with the corresponding local directions of the anatomical axes. This dataset was used for the training and validation of a deep neural network, ResNet-101, featuring a top layer having a linear activation function with 9 outputs aimed at the estimation of the three dimensional components of the three axes. The accuracy of the network was qualitatively (Fig. 1) and quantitatively tested on a large group of simulated radiographic images for which the direction of the axes was manually measured in the original CT dataset, resulting in absolute errors in the range of 1 to 5 degrees. The novel method will be useful to extract three dimensional information about the spinal alignment from planar images even in clinical cases in which vertebrae can be markedly rotated due to spinal deformities or to an imprecise alignment of the patient with respect to the detector.

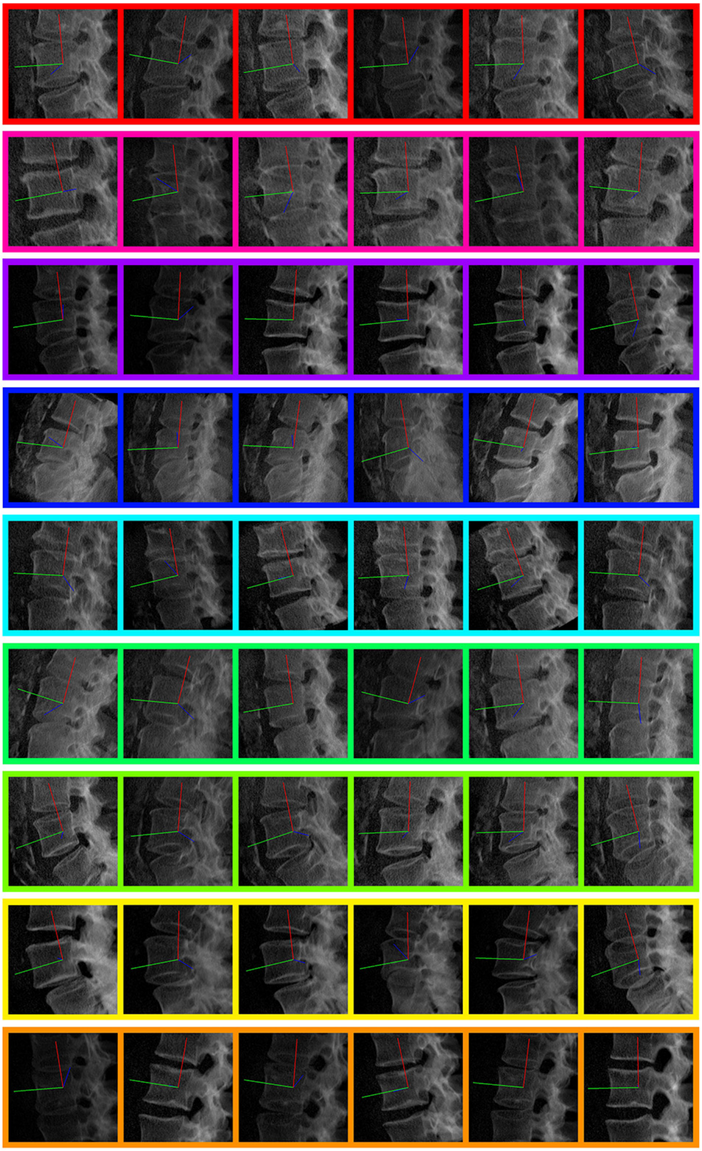

Fig. 1: Examples of three dimensional anatomical axes extracted from planar radiographs of 9 vertebrae. Each row represents 6 different views of the same vertebra constructed from CT scans. The anteroposterior axis is shown in green, the craniocaudal axis in red, whereas the laterolateral axis is in blue.

Which Landmark is Best Suited to Assess the Thoracic Orientation?

Zander T, Pan F, Schmidt H

Julius Wolff Institute, Charité – Universitätsmedizin Berlin, Berlin, Germany

The need of accurate spinal kinematic measurements for clinical and research purposes is undisputed. Several skin surface based techniques exist to measure the position and orientation of bony structures [1]. The accuracy of these techniques are, however, limited by errors associated with the movements of the interface between the skin and the underlying rigid structure whose motion is to be measured. Furthermore, structures like the thorax are frequently regarded substantially rigid but show considerable mobility within itself [2]. This study was aimed to quantify this mobility at different thoracic landmarks in young healthy subjects during functional activity to give a recommendation for the best suited measurement location.

The locations of 29 landmarks were continuously captured on currently six (later on for the workshop 20) subjects (age: 26–56 yrs.) during sitting, standing, walking, jumping, and different breathing depths using reflective markers (Vicon, Oxford, GB). Since single markers do not provide information about local orientations, marker triplets were used at every landmark (Fig. 1). For every time frame, local rotations were then determined by first backtracking the rigid body motion (RBM) of the thorax in general, and subsequently calculating the RBM of each rigid marker triplet [3]. The latter one was finally converted to axis angles which denote the measurement error at a particular landmark.

Landmarks at the lower end of the ribcage showed the largest errors (24°, Fig. 2). However, the inter-subject variability was large. Landmarks at the cranial sternal region (especially at *Louis angle*) and at the T3 spinous process showed the smallest errors (<4° and <5°, respectively) for nearly all subjects and tasks. Normal breathing alone led to a error of at least 1°.

It is therefore recommended to use the cranial sternal region to assess the thoracic orientation. Errors in the order of 3° can, however, not be avoided as long as no additional measuring points are used.

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[3] Besl, JP and McKay, ND: Method for registration of 3-D shapes, Proc. SPIE 1611, 1992; doi: 10.1117/12.57955

Fig. 1: Marker triplet to measure skin orientations.

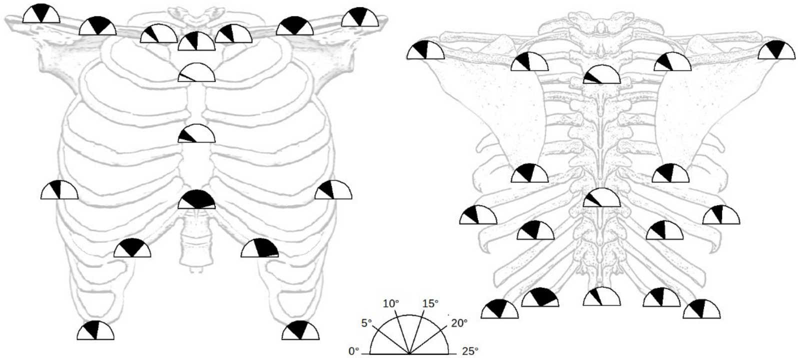

Fig. 2: Expected range (subject variability) of maximum (exercise variability) errors on ventral (left) and dorsal side (right) of the thorax.

In-Vivo Hip and Lumbar Spine Implant Loads during Activities in Forward Bent Postures

Damm P, Reitmaier S, Hahn S, Duda G, Schmidt H

Julius Wolff Institute, Charité - Universitätsmedizin Berlin, Berlin, Germany

Long-term measurements on the lumbar spinal alignment during daily life revealed that humans spent 90% of the day in a forward bent posture [1]. Compared to upright standing, this posture leads to a substantial increase in spinal loading [2-4]. Own kinematic investigations proved that during flexion, the lumbar spine and pelvis contribute differently to the total amount of motion [5]. These variances in the temporal kinematics of the spine and the hip, however, might correlate with a different occurrence of the maximum loads within both structures during flexion. For a sophisticated understanding of the load distribution between the hip and the spine, this paper aims to evaluate the *in vivo* contact forces within hip and spine implants during activities in flexed bent postures.

This work exploited data collected in earlier *in vivo* measurements on 10 patients with telemeterized hip endoprostheses (HE) and 5 with vertebral body replacements (VBR). The following activities were investigated: pure upper body flexion, dropping and lifting of 10 kg weights with straight and bent knees, sitting down and standing up.

The maximum forces in VBR were considerably lower than in HE (Fig. 1). Increases in pure upper body flexion lead to direct increases of the resultant forces within VBR, followed by a plateau or even a decrease of the force at an inclination angle of approximately 33°. The resultant force in HE started to increase at a later phase of inclination and passed into an almost continuous increase until the maximum inclination. This general curve behavior was only slightly influenced by carrying additional weights during the tasks or different lifting techniques (stoop vs. squat). However, the measured resultant forces differed between the hip and the spine. Whereas only a small difference (4%) was observed in the measured resultant force between stoop and squat lifting in VBR, a large difference (19%) was found in HE.

Results emphasize that maximum loads in the anterior spinal column not necessarily occur at maximum upper body inclination as usually expected, but already at intermediate flexion angles in VBR patients. In contrast, maximum loads in HE actually occur at maximum inclination angles. These findings are in agreement with lumbo-pelvic ratio measurements, where the greatest contributions for upper body flexion of the lumbar spine occurred in the first phase of flexion followed by lower contributions in later phases of flexion [5].

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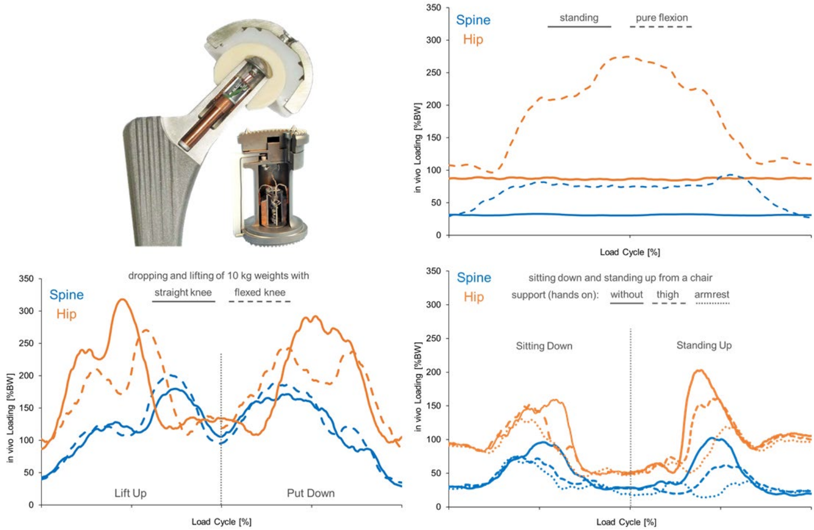


Fig. 1: Instrumented spine and hip replacements and their in vivo measured loads during activities in forward bent postures.

Bottom-up versus Top-down L5/S1 Moment Estimation during Manual Lifting using an Ambulatory Measurement System

Faber GS[a,b,c], Chang CC[c,d], Kingma I[a], Dennerlein JT[b,e], van Dieën JH[a]

[a]*Dept. of Human Movement Sciences, Faculty of Behavioural and Movement Sciences, Vrije Universiteit Amsterdam, Amsterdam Movement Sciences, Amsterdam, The Netherlands*
[b]*Dept. of Environmental Health, Harvard T.H. Chan School of Public Health, Boston, MA, USA*
[c]*Liberty Mutual Research Institute for Safety, Hopkinton, MA, USA*
[d]*Dept. of Industrial Engineering & Engineering Management National Tsing Hua University, Taiwan, ROC*
[e]*Dept. of Physical Therapy, Movement and Rehabilitation Sciences, Northeastern University, Boston, MA, USA*

Manual lifting can result in high low-back loading [1] which is probably the main reason why it is also an important risk factor for occupational low-back pain [2]. Therefore, many laboratory studies have investigated low-back loading during manual lifting, using inverse dynamics to calculate L5/S1 joint moments. These L5/S1 moments can be calculated using either a bottom-up (based on force plate data plus lower body kinematics) or a top-down (based on hand forces plus upper body kinematics) approach [3].

More recently, for measurements in the field, wearable inertial motion capture (IMC) and Force Shoe (FS) systems have been developed to obtain segment kinematics and GRFs, respectively. In the current study, we utilized these ambulatory measurement systems to calculate the L5/S1 moments using a bottom-up and top-down approach. For practical reasons, rather than measuring hand forces (HFs), these were estimated based on FS data and full body kinematics, which has previously shown to work well [4]. As a gold standard reference, a laboratory-based bottom-up model was used (LAB_{bottom}) [3].

Eight male participants lifted a 10-kg box from ground level (Fig. 1A), while 3D full-body kinematics were measured using an optical motion capture (OMC, Optotrak) and an IMC (MVN, Xsens) system, and 3D GRFs were measured using a force plates (FPs, Kistler) and FSs (ATI/Xsens). L5/S1 moments were calculated 3 times based on different data sources:

1) FP	+ OMC_{lower}	(LAB_{bottom}, gold standard reference),
2) FS	+ IMC_{lower}	($AMBU_{bottom}$),
3) HF	+ IMC_{upper}	($AMBU_{top}$),

As a measure of system performance, RMS errors were calculated between reference LAB_{bottom} moments on one hand and the ambulatory ($AMBU_{bottom}$ & $AMBU_{top}$) moments on the other hand.

The results (Fig. 1E) show that the $AMBU_{top}$ performed much better (averaged over subjects, RMS errors up to 15Nm) than the $AMBU_{bottom}$ system (averaged over subjects, RMS errors up to 49Nm). The reason for this is that, for the bottom-up approach, small position errors at the feet result in large L5/S1 moment errors because of the high GRFs (Fig. 1D).

In conclusion, for ambulatory L5/S1 moment assessment with an IMC+FS system, using a top-down inverse dynamics approach with estimated hand forces is much more accurate than a bottom-up approach.

[1] Faber GS et al. (2009). Working height, block mass and one- vs. two-handed block handling: the contribution to low back and shoulder loading during masonry work. Ergonomics. 52: 1104-18. [2] Coenen et al. (2014). Cumulative mechanical low-back load at work is a determinant of low-back pain. Occup Environ Med 71, 332-337. [3] Kingma I et al. (1996).Validation of a full body 3-D dynamic linked segment model. Human Movement Science 15, 833–860. [4] Faber GS et al. (2018). Continuous ambulatory hand force monitoring during manual materials handling using instrumented force shoes and an inertial motion capture suit. Journal of Biomechanics, 46, 2736–2740.

Acknowledgements: This work was supported by the European Union's Horizon 2020 through the SPEXOR project, contract no. 687662.

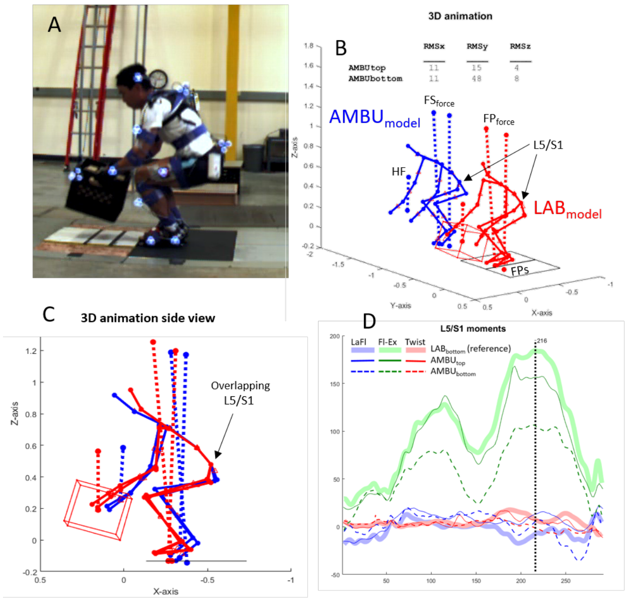
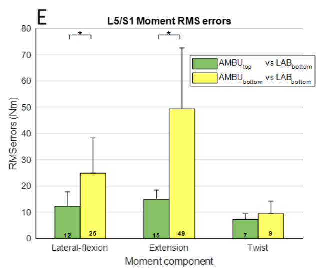


Fig. 1:
A: Subject during the experiment.
B: Matlab visualization of the 3D model.
C: Side view of the Matlab visualization of 3D model.
D: Typical example of the 3D moment curves.
E: L5/S1 moment RMS errors, averaged over 8 subjects.



A Prospective Study of Lumbo-pelvic Coordination in Patients with Non-chronic Low Back Pain

Shojaei I[a], Salt EG[b], Bazrgari B[a]

[a]*F. Joseph Halcomb III, M.D. Dept. of Biomedical Engineering, Lexington, KY, USA*
[b]*College of Nursing, University of Kentucky, Lexington, KY, USA*

Despite the current understanding describing abnormalities in the lumbo-pelvic coordination of patients with non-specific low back pain (LBP) [1], it remains unclear how such abnormalities change with time.

The changes in timing and magnitude aspects of lumbo-pelvic coordination, during a trunk forward bending and backward return task, along with the alterations in pain intensity and disability level were investigated over a six-month period in 29 patients who had non-chronic LBP at the time of enrollment in the study. To enable investigation of baseline abnormalities in lumbo-pelvic coordination of patients, we also included lumbo-pelvic coordination data of age and gender-matched back healthy individuals from an earlier study of our group [2, 3]. Moreover, we investigated the differences in lumbo-pelvic coordination between patients with moderate-severe LBP (i.e., those whose level of pain was ≥ 4 (out of 10) at all three data collection sessions; n=8) and patients with low-moderate LBP (n=21).

The abnormal lumbo-pelvic coordination of patients with non-specific LBP, observed at baseline, persisted over the course of study period despite significant reduction in their pain intensity (>18%) and disability level (>10%) (Figs. 1 and 2). There were clear distinctions in measures of lumbo-pelvic coordination between patients with low-moderate and moderate-severe LBP (Fig. 1). Contrary to our expectation, however, the abnormalities in timing and magnitude aspects of lumbo-pelvic coordination, particularly under fast-paced tasks, were larger in patients with low-moderate LBP (Fig. 1).

Distinct but lingering abnormalities in lumbo-pelvic coordination, observed in patients with low-moderate and moderate-severe LBP, might have a role in persistence and/or recurrence of symptoms in these patients. Such inferences, however, should further be studied in future via investigation of the relationship between abnormalities in lumbo-pelvic coordination and clinical presentation of LBP.

[1] Laird et al 2014, BMC Musculoskeletal Disorder 15, 1.
[2] Shojae et al 2016, Journal of Biomechanics 49, 896-903.
[3] Vazirian et al 2017, Ergonomics 60, 967-76

Acknowledgment: This work was supported in part by the National Center for Research Resources and the National Center for Advancing Translational Sciences (UL1TR000117) as well as an award (R21OH010195) from the Centers for Disease Control and Prevention. The content of this manuscript is solely the responsibility of the authors and does not necessarily represent the official views of the National Institute of Health and the Centers for Disease Control and Prevention.

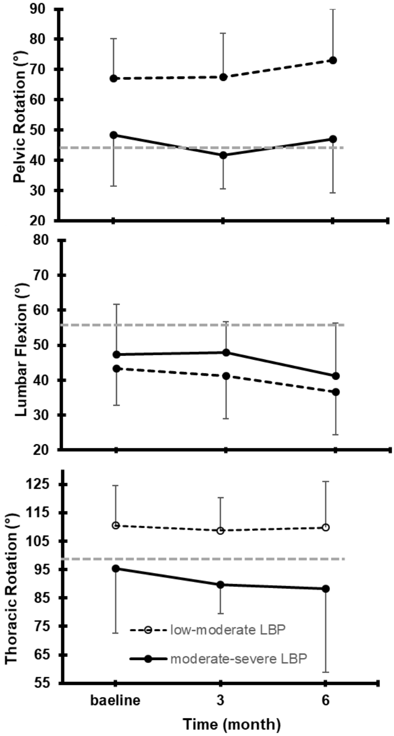


Fig. 1: The magnitude of pelvic and thoracic rotations and lumbar flexion at the end of bending phase. Gray broken lines represent the mean values from controls. Data collections were conducted immediately after enrolment (baseline), ~3 and ~6-month post-enrolment.

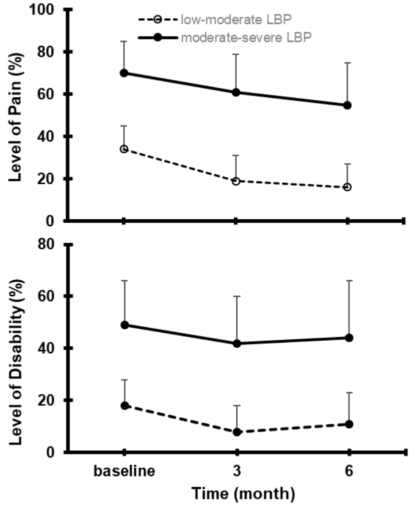


Fig. 2: Alterations in measures of pain (top) and disability (bottom) over the study period. Both pain and disability levels are presented as percent of maximum possible value (i.e., 10 for pain level according to the Wisconsin Brief Pain Inventory and 24 for disability level according to the Roland Morris Disability Scale).

Patient-Specific Changes in Adjacent Segment Kinematics After Lumbar Decompression and Fusion

Wawrose R[a], Byrapogu V[a], LeVasseur C[a], Shaw JD[a], Donaldson W[a], Lee J[a], Aiyangar A[a,b], Anderst W[a]

[a]*Dept. of Orthopedic Surgery, University of Pittsburgh, Pittsburgh, PA, USA*
[b]*Swiss Federal Laboratories for Materials Science and Technology, Duebendorf, Switzerland*

The pathogenesis of adjacent segment disease is thought to be secondary to altered biomechanics resulting from fusion [1]. Direct in vivo evidence for altered biomechanics following lumbar fusion is lacking. The clinical standard for measuring lumbar motion is static end-range flexion and extension lateral radiographs [2] which are not able to assess the midrange motion that comprises most activities of daily living. This study's aim was to determine the effects of lumbar fusion on in vivo adjacent segment kinematics over the entire dynamic flexion activity. We hypothesized that flexion and AP translation of the superior adjacent segment would increase post-fusion.

Seven patients with symptomatic lumbar degenerative spondylolisthesis (5 M, 2 F; age 65±5.1 years) stood within a biplane radiographic imaging system and performed two to three trials of continuous flexion of their torso [3] while synchronized biplane radiographs were acquired at 20 images per second. Testing occurred one month before (PRE) and six months after (POST) lumbar spinal decompression and fusion surgery. A validated volumetric model-based tracking process was used to track the position and orientation of vertebrae in the radiographic images [3]. Intervertebral flexion/extension and AP translation (slip) at the superior adjacent segment were calculated over the entire dynamic flexion activity. Skin-surface markers were tracked using conventional motion analysis to determine torso flexion. PRE to POST differences were considered measurable if they were more than twice the validated uncertainty in our measurement system (0.5° for flexion/extension and 0.2 mm for slip) [3].

There were no consistent trends in the observed changes in superior adjacent segment kinematics after lumbar fusion (Fig. 1).

Additional research is warranted to identify factors that predict patient-specific changes in adjacent segment motion after fusion. These in vivo results contradict in vitro testing that suggests adjacent segment motion increases after fusion.

[1] Tobert et al., Clin Spine Surg, 2017.
[2] Axelsson et al., Spine 1997.
[3] Dombrowski et al., Eur Spine J, 2018.

Acknowledgements: This work was supported by Swiss National Science Foundation (SNSF) Ambizione Career Grant PZ00P2154855/1 and NIH grant 5R44AR064620.

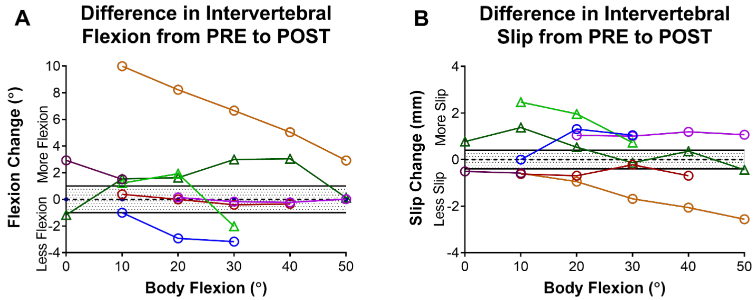


Fig. 1: Changes in adjacent segment flexion (A) and slip (B) from PRE to POST at 10° incriments of body flexion for seven DS patients. The shaded area represents the precision of our measurement. Values outside these boundaries represent measureable changes from PRE to POST. Triangles identify L2/L3 as the adjacent segment while circles identify L3/L4 as the adjacent segment.

The Impact of Curve Severity on the Pelvic Kinematic and Erector Spinea and Gluteusmedius Muscles Activity during Gait in Patients with Adolescent Idiopathic Scoliosis

Yazdani S[a], Farahpour N[b], Allard P[c]

[a]*Dept. of Motor Control, Faculty of Physical Education and Sport Science, University of Tabriz, Tabriz, Iran*

[b]*Kinesiology Dept., Bu Ali Sina University, Hamedan, Iran*

[c]*Kinesiology Dept., Montréal University, Montréal, Canada*

Adolescent idiopathic scoliosis (AIS) is a complex three dimensional deformity of the spine. In some cases (~10%) the curvature becomes progressive and may reach above 40-50° before the end of the growth. This study aimed to investigate the influence of curve severity on the pelvic kinematic and erector spinea and gluteusmedius muscles activity during gait in AIS patients.

Twenty AIS patients with right thoracic curvature (10 severe and 10 mild scoliosis) and 10 healthy control subjects were studied. Using a VICON 460 motion analysis system and a surface EMG system the lower limbs' kinematics and the activity of bilateral PS at T6 (PST6), T10 (PST10), and L3 (PSL3) levels and GM muscles were measured during walking.

Results showed that the EMG of the right PST6 and bilateral GM muscles in both scoliosis groups were higher than those in control group. The muscle activity alterations were similar in both scoliosis groups. Also, AIS patients demonstrated asymmetrical PET6 activity during walking. The mean left pelvic rotation in sever and mild scoliosis groups were about 27% ($p=0.04$) and 15% ($p=0.02$), respectively, greater than that in control group. In sever scoliosis group, the mean left pelvic obliquity was significantly greater than that in control group ($p=0.02$). Both experimental groups showed asymmetrical pelvic range of motion.

Walking in both severs and mild AIS patients is associated with higher and asymmetrical activity of the right PST6 and GM muscles and pelvic kinematics. The muscle activity and kinematic alterations were similar in both scoliosis groups. Severe AIS is associated with higher pelvic obliquity. During scoliosis rehabilitation, an attention should be paid to patients' hip balance, and selective strengthening of the PS muscles. Also, further longitudinal study is recommended to address the link between the curve progression and kinematic components.

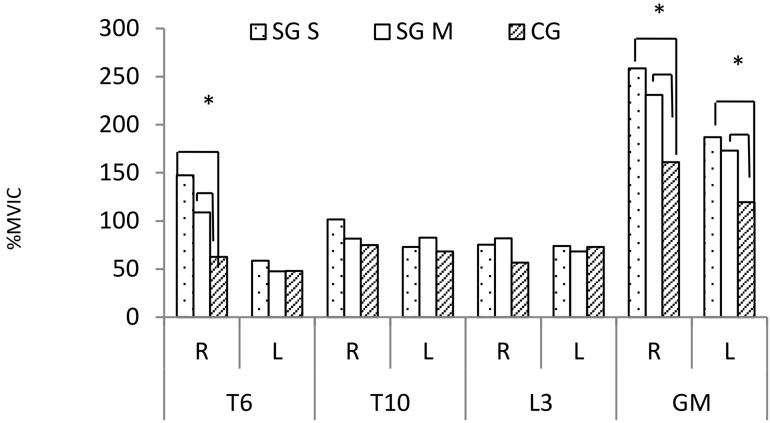


Fig. 1: Normalized EMG activity of erector spinea and gluteusmedius muscles. (SG S: sever scoliosis group, SG M: mild scoliosis group, CG: control group).

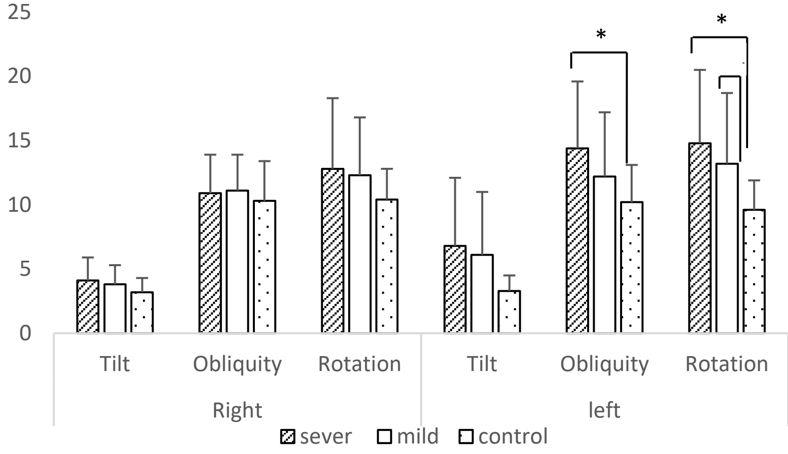


Fig. 2: Comparison of pelvic kinematics in three different planes between sever SG, mild SG and CG.

Automatic Generation of Patient-Specific FE Models of the Lumbar Spine

Caprara S[a,b], Senteler M[a,b], Snedeker JG[a,b], Farshad M[a]

[a]*Dept. of Orthopaedics, Balgrist Hospital, University of Zurich, Zurich, Switzerland*
[b]*Institute for Biomechanics, Swiss Federal Institute of Technology (ETH), Zurich, Switzerland*

Finite Element (FE) models of lumbar functional spinal units (FSU) are a promising tool to optimize spinal fusion parameters by accounting for bone and implant loads under consideration of spinal biomechanical behavior. Although efforts have been pursued to automatically generate population-based FE models [1], the application of FE analysis in clinical practice still lacks the level of automation required for high throughput application and generic models are not suitable for patient-specific surgical planning. Hence the aim of the present study is to create a pipeline to automatically build patient-specific FSU FE models.

For each lumbar vertebra a Statistical Shape Model (SSM) was generated using the open-source Scalismo package [2]. The level-specific SSMs were trained with the following numbers of segmented vertebrae obtained from clinical CTs acquired at Uniklink Balgrist (Zurich): L1: 71, L2: 100, L3: 138, L4: 110, L5: 77.

The resulting models were used as input for a custom-built FE model generator. The FSU FE models are generated for the FEBio open-source FE software, based on the recently published open-access model of the human lumbar spine [3]. The correspondence property of the non-rigid registration of the SSMs along with algorithmic analysis of mesh element normals enabled the automated detection of endplates as well as the localization of facet joints and ligament attachment sites.

The pipeline outputs a FEBio FE model ready for simulation. The pipeline is able to automatically create models of L1-L2, L2-L3, L3-L4, and L4-L5 lumbar FSUs and simulate flexion, extension, lateral bending, and axial rotation by applying a combined load (moment, compression) at the upper vertebral endplate.

The resulting SSMs were evaluated using a leave-one-out (LOO) experiment, the RMS errors between registered and manually segmented vertebra for all the five lumbar levels are presented in Fig. 1. Both the contour plot and the statistical analysis revealed errors between 0.5 and 2.6 mm, which are comparable to those resulting from different manual segmentations. Fig. 1 shows how the pedicle and spinous process regions are the most challenging to fit presenting higher RMS average error compared to the error at the vertebral body.

The proposed pipeline integrates two different open-source packages to automatize the creation of patient-specific FE models of various FSUs. Preliminary simulation results show a good agreement with literature data and model outputs reported in [3]. The differences between FE simulations created by the presented pipeline and manually generated FE models are currently investigated in detail. Furthermore, we aim at incorporating open source Deep Learning techniques to fully automatize vertebra localization in clinical CT datasets and the non-rigid registration of the SSMs.

By incorporating open source modeling and analysis packages we strongly contribute to the need of open-access modeling pipelines for improved reproducibility and verification of results in patient-specific modelling.

[1] Campbell et. al; Automated finite element meshing of the lumbar spine: Verification and validation with 18 specimen-specific models; J Biomech, 2016

[2] Clogenson et. al; A Statistical Shape Model of the Human Second Cervical Vertebra; Int J Comput Assist Radiol Surg, 2014

[3] Finley et. al; FEBio finite element models of the human lumbar spine; Comput Methods Biomech Biomed Engin, 2018

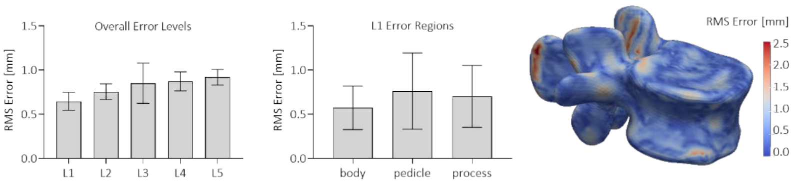


Fig. 1: Resulting errors in mm from the iterative LOO experiment performed on all the vertebral models using all the training vertebrae.
Error distribution [mm] on a patient-specific L2 vertebra fitted using the corresponding SSM. It is visible how the larger errors are present at the spinous processes position.

Effect of a Passive Exoskeleton on Mechanical Loading during Dynamic Lifting

Koopman AS[a], Kingma I[a], de Looze MP[b], van Dieën JH[a]

[a]*Dept. of Human Movement Sciences, Faculty of Behavioural and Movement Sciences, Vrije Universiteit Amsterdam, Amsterdam Movement Sciences, Amsterdam, The Netherlands*
[b]*TNO, Leiden, The Netherlands*

Spine loading during dynamic lifting has been shown to be a risk factor for the development of low back pain [1]. Assistive devices are being developed with the aim to reduce spine compression during dynamic lifting. The aim of this study was to investigate the effects on spine moments of two variants of a passive lifting device.

Eleven healthy participants lifted a 10kg box from ankle and knee height, with a free style lifting technique. Lifts were performed without a device, and with two types (LOW and HIGH) of the device (Laevo BV, The Netherlands), which generated peak moments at large and moderate flexion angles, respectively. Full-body kinematics, ground reaction forces and back and abdominal muscle EMG were measured. Based on device angles, device moments were subtracted from the net moment calculated with bottom-up inverse dynamics to estimate the subject generated L5S1 moment.

Due to hysteresis, the devices provided more support during downward motions (about 25Nm) than in upward motions (about 15Nm) where peak extension moments were seen. When lifting at ankle height, in contrast to expectations, resultant peak moments were significantly reduced (by on average 19Nm) with the HIGH but not with the LOW device (Fig. 1). This was due to some subjects reaching the hard end-stop of the HIGH device, thereby increasing its support. The lack of effect of the LOW device was due to subtle changes in lifting behaviour. For lifts at knee height, the HIGH device did not significantly reduce peak moments because it was already beyond its maximum support range, whereas the LOW device reduced peak moments by on average 17Nm.

In conclusion, the support devices were less effective in upward than in downward motions, which limits their effects during lifting. Reduction of hysteresis and stronger springs could improve the effect of the devices.

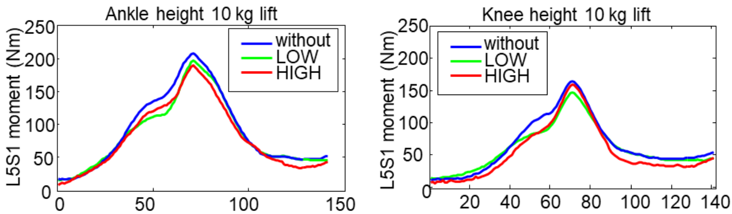


Fig. 1: Effect of two support devices on L5S1 extension moments during ankle (left) and knee (right) height lifts, averaged over participants (N=11).

[1] Coenen P. et al., Occupational and Environmental Medicine 71, 332-337 (2014).

Acknowledgements:

This work was supported by the European Union's Horizon 2020 through the SPEXOR project, contract no. 687662. We would like to thank Laevo B.V. for providing the two devices.

Sex-Dependant Estimation of Spinal Loads during Static Manual Material Handling Activities - Combined In-Vivo and In-Silico Analyses

Firouzabadi A[a], Pan F[a], Zander T[a], Arjmand N[b], Schmidt H[a]

[a]*Julius Wolff Institute, Charité – Universitätsmedizin Berlin, Berlin, Germany*
[b]*Dept. of Mechanical Engineering, Sharif University of Technology, Tehran, Iran*

Manual material handling (MMH) is one of the prime causes of low back injuries [1]. To assess the risk of biomechanical injuries, majority of previous studies have evaluated spinal loads [2] while mainly focusing on male individuals. However, females perfom MMH tasks using different postural techniques and muscle recruitments [3]. Therefore, in order to properly investigate the effects of inter-sex and individual differences on spinal loads, one needs to use subject-specific musculuskeltal models that are also driven by subject-specific *in vivo* kinematics data.

Kinematics of 45 skin markers (Vicon, Oxford, UK) and ground reaction data were collected from twenty volunteers (10 males and 10 females) during static standing, peak voluntary flexion, and eleven different one- or two-handed symmetric and asymmetric lifting (10 kg) tasks (Fig. 1) [4]. Simultaneously, and for validation purposes, electromyographic (EMG) data were recorded from 12 abdominal and back muscles. Kinematics and ground reaction forces were input into an inverse dynamic musculoskeletal model (Anybody v.6.1, Aalborg, Denmark) that estimated muscle forces and spinal forces via an optimization algorithm. The model was scaled with respect to subject height, weight, and segment lengths according to the position of the reflective markers [5].

Priminarly results (from 2 males and 2 females) indicate that the largest inter-individual differences in lumbar segmental compressive forces were ~284% of body weight (in task 12). The highest normalized loads (to BW) occurred for a female subject holding asymmetricly 10 kg (two-handed) (i.e., task 12). Normalized L5-S1 compression loads to BW were generally higher for females (Fig. 2).

Current results suggest that females are exposed to higher spinal loads (relative to their BW) than males when performing identical lifting tasks and thus females may be at higher risk of low back injuries. For preventive intervention and risk assessments, sex- and inert-subject differences should hence be considered.

[1] Davis, K. G. J., Michael J. (2005). "Biomechanical modeling for understanding of low back injuries: A systematic review." Occupational Ergonomics 5(1): 57-76.
[2] Parida, R. and P. K. Ray (2015). "Biomechanical Modelling of Manual Material Handling Tasks: A Comprehensive Review." Procedia Manufacturing 3: 4598-4605.
[3] Plamondon, A., et al. (2017). "Difference between male and female workers lifting the same relative load when palletizing boxes." Appl Ergon. Apr;60:93-102.
[4] Rajaee, M. A., et al. (2015). "Comparative evaluation of six quantitative lifting tools to estimate spine loads during static activities." Applied Ergonomics 48: 22-32.
[5] Andersen, M. S., et al. (2010). "A computationally efficient optimisation-based method for parameter identification of kinematically determinate and over-determinate biomechanical systems." Comput Methods Biomech Biomed Engin. 13(2):171-83.

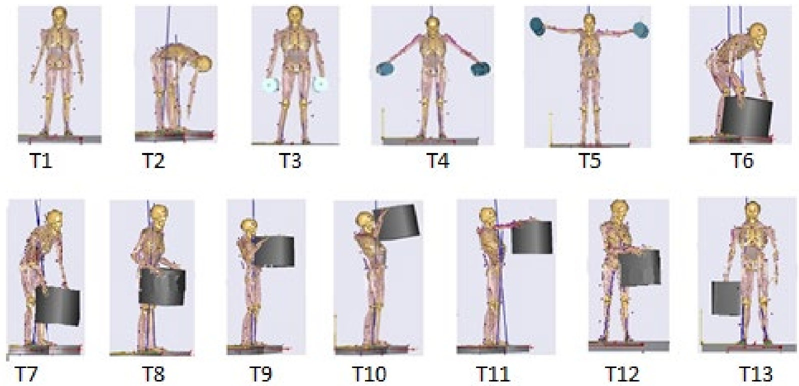


Fig. 1: Standing, flexion and 11 different loaded lifting tasks (T3 - T13) performed by each participant.

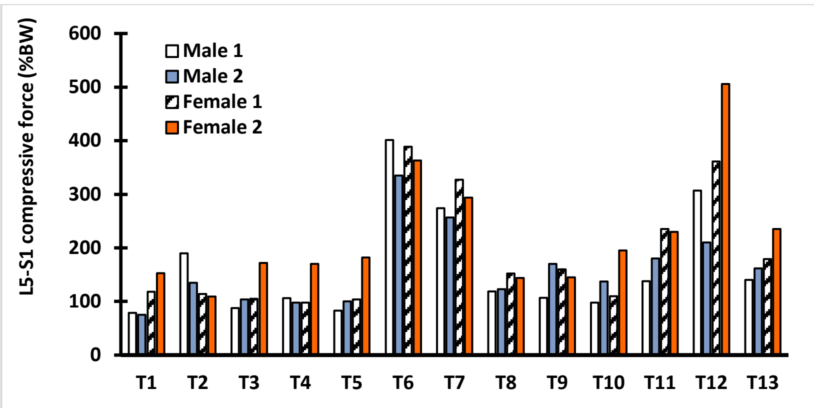


Fig. 2: Predicted compressive force on L5-S1 joint during different lifting tasks as percentage of body weight (%BW).

Subject-Specific Regression Equations to Estimate Spinal Loads in Asymmetric Static Lifting

Ghezelbash F[a], Shirazi-Adl A[a], El-Ouaaid Z[a], Plamondon A[b], Arjmand N[c]

[a]*Polytechnique Montréal, Québec, Canada*
[b]*IRSST, Québec, Canada*
[c]*Sharif University of Technology, Tehran, Iran*

Excessive spinal load is recognized as a risk factor of back pain. To estimate spinal loads, measurements are indirect and/or invasive (e.g., intradiscal pressure – IDP) whereas musculoskeletal (MS) modeling is a commonly used alternative. However, due to the complexity of MS models, regression equations are developed as easy-to-use tools. Existing regression equations are not personalized and mostly limited to symmetric lifting tasks. Thus, we aim to develop subject-specific regression equations to estimate spinal loads during asymmetric static lifting tasks using our MS model and concurrent in vivo kinematics-EMG measurements.

With institutional approval and consent, we recorded whole body kinematics and EMG activities of abdominal and back muscles in 9 female and 10 male healthy individuals as they performed 64 symmetric and asymmetric static lifting tasks with different hand-loads (2-14 kg). We simulated foregoing and additional tasks using our subject-specific nonlinear finite element MS model of trunk [1] driven by measured kinematics for different body heights, body weights and sex. Quadratic regression equations (inputs: trunk flexion, asymmetry angle, load magnitude, load lever-arm, body height, body weight, sex; Fig. 1) were developed to estimate L4-L5 and L5-S1 shear and compression loads. For validation, estimated muscle activities and L4-L5 IDPs were compared with our own EMG and reported in vivo IDP measurements.

Low average absolute error (<9%) and high correlation coefficient (>0.97) demonstrated satisfactory goodness-of-fit of regression equations. Trunk flexion angle, asymmetry angle, hand-load weight, hand-load moment arm and body weight contributed the most to spinal loads (Fig. 1c). Estimated muscle activities had moderate agreement with our measured EMGs while predicted L4-L5 IDPs were in strong agreement with measurements (R2=0.85; Fig. 2).

With both the MS model and regression equations adjusted in accordance with subjects' anthropometric parameters (in EMG and IDP experiments), satisfactory agreement demonstrates the relative accuracy of the model and regression equations when estimating spinal loads during asymmetric tasks. Proposed equations can be used as an accurate tool when evaluating spinal loads in various occupational tasks.

[1] Ghezelbash F, et al. (2016) Biomech Model Mechanobiol. 15:1699-17

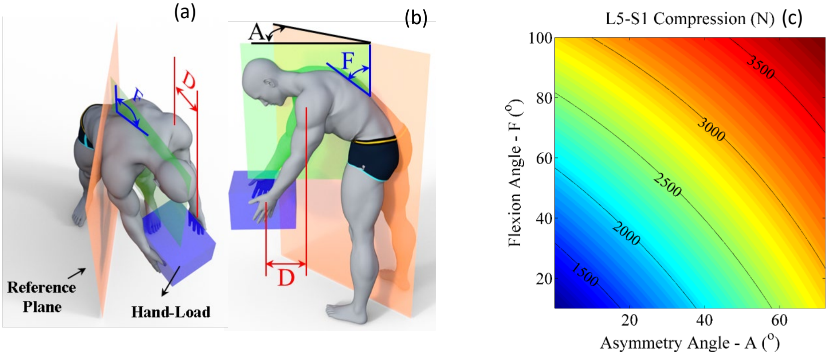


Fig. 1: (a, b) Schematic representation of an asymmetric lifting task (A: asymmetry angel; F: fexion angle; D: moment arm from the shoulder joint), and (c) contour plot of L5-S1 compression (N) computed from regression equations at hand-load weight=10 kg, D=0 cm, sex=male, body height=175 cm and body mass=75 kg.

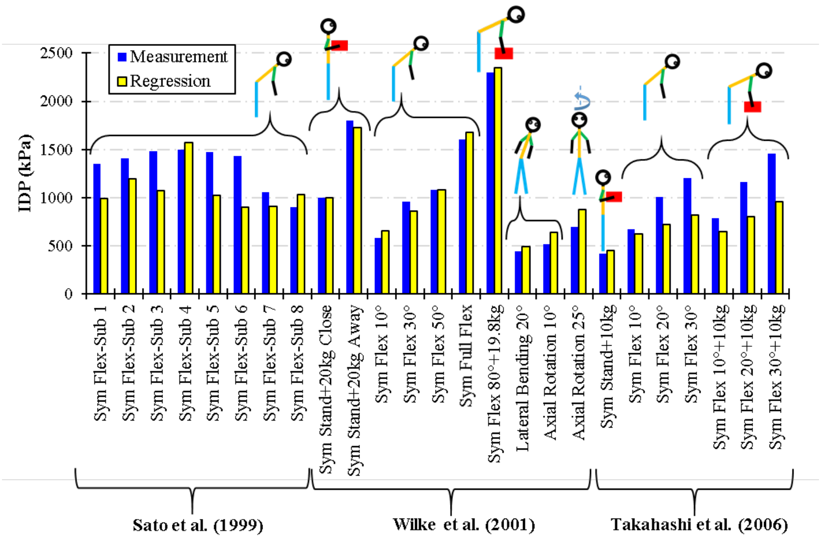


Fig. 2: Estimated subject-specific IDPs versus reported in vivo measurements during various symmetric and asymmetric tasks (Flex: Flexion; Sub: Subject; Sym: Symmetric).

Sensitivity of Musculoskeletal Model-based Lumbar Spinal Loading Estimates to Type of Kinematic Input and Passive Stiffness Properties

Byrne RM[a], Zhang X[b], Aiyangar AK[c,d]

[a] *Dept. of Mechanical Engineering, University of Pittsburgh, Pittsburgh, PA, USA*
[b] *Dept. of Industrial & Systems Engineering, Texas A&M University, College Station, TX, USA*
[c] *Mechanical Systems Engineering, EMPA-Swiss Federal Laboratories for Materials Science and Technology, Duebendorf, Switzerland*
[d] *Dept. of Orthopaedic Surgery, University of Pittsburgh, Pittsburgh, PA, USA*

Computational models offer the only viable non-invasive possibility for assessing lumbar loads, but outputs can be sensitive to the accuracy of input kinematics and passive stiffnesses.

A generalized OpenSim full-body musculoskeletal model was constructed by combining: 1) lower-body model by Arnold et al. [1] and 2) upper body model by Senteler et al. [2]. A healthy female participant (25 years, 61kg) performed dynamic lifting and upright standing tasks while 6DOF lumbar kinematics and whole-body surface marker-based kinematics were captured by a Dynamic Stereo-radiography (DSX) system and an 8-camera Vicon motion capture system respectively [3]. The generalized model was scaled to the participant, and the net joint moments (NJM) and joint reaction forces (JRFs) at the L4L5 were computed.

In a stepwise manner, changes were made to the lumbar spine portion to observe how NJMs and JRFs were affected by four factors (24 different models):

1) Lumbar kinematics: (a)Rhythmic distribution without translations (b) DSX-based 6DOF kinematics)
2) Upright standing preloads (a) initial compression or (b) no initial compression)
3) Passive disc stiffness (a) No passive stiffness; (b) linear stiffness; (c) nonlinear stiffness)
4) External weight lifted (10lb or 30lb)

Increasing the weight lifted led to increased maximum JRFs in all models, but with substantially larger increases in models implementing DSX-based kinematics (123%-306% vs. 44%- 47% for rhythmic distribution). The degree to which kinematic input affected L4L5 JRFs depended on whether passive bushing elements were included, and on the linearity or nonlinearity of the stiffnesses (Fig. 1). For DSX-based kinematics, the bushing-dependent variation in JRFs was much greater compared to the input of rhythmic distribution: e.g. compressive JRF reduced by more than half between linear and nonlinear bushings with upright pre-load implementation.

While inclusion of in vivo data into lumbar models may potentially provide improved loading estimates, interaction with other input parameters must be taken into account.

[1] Arnold, E. M., Ward, S. R., Lieber, R. L., and Delp, S. L., 2010, "A model of the lower limb for analysis of

human movement," Ann Biomed Eng, 38(2), pp. 269-279.

[2] Senteler, M., Weisse, B., Rothenfluh, D. A., and Snedeker, J. G., 2015, "Intervertebral reaction force prediction using an enhanced assembly of OpenSim models," Computer methods in biomechanics and biomedical engineering, pp. 1-11.

[3] Aiyangar, A. K., Zheng, L. Y., Tashman, S., Anderst, W. J., and Zhang, X. D., 2014, "Capturing Three-Dimensional In Vivo Lumbar Intervertebral Joint Kinematics Using Dynamic Stereo-X-Ray Imaging," J Biomech Eng-T Asme, 136(1), p. 011004.

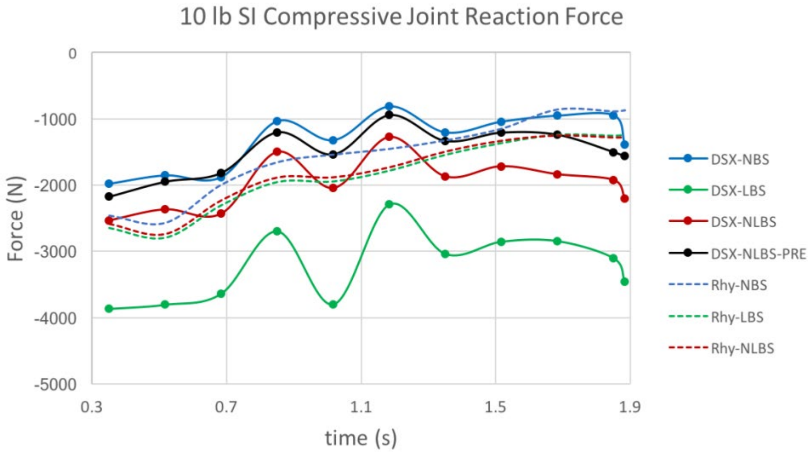


Fig. 1: Superior-inferior (SI) compressive joint reaction forces for different variations of the lumbar spine during sagittally symmetric lifting task with the 10 lb (4.5 kg) external weight. NBS = no bushings, LBS = linear bushings, NLBS = nonlinear bushings, PRE = preload.

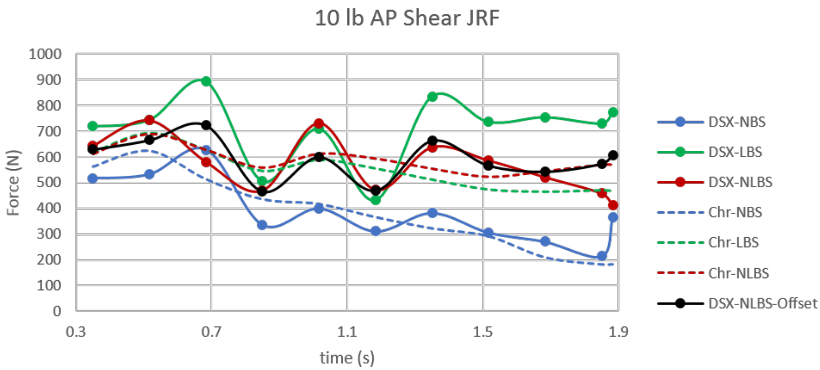


Fig. 2: Anterior-posterior (AP) shear joint reaction forces for different of the lumbar spine during sagittally symmetric lifting task with the 10 lb (4.5 kg) external weight. NBS = no bushings, LBS = linear bushings, NLBS = nonlinear bushings, PRE = preload.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Assessment of Spine Loading via a 2 Muscle Model vs. 10 Muscle Model during One vs. Two Handed Lifting Tasks

Weston E, Aurand A, Dufour J, Knapik G, Marras WS

Spine Research Institute, Dept. of Integrated Systems Engineering, The Ohio State University, Columbus, Ohio, USA

Accurate evaluation of spine loads is critical for the understanding of causal pathways leading to low back disorders since tissue force is the stimulus for the pain cascade. The objective of this study was to assess the impact of modeling two styles of lifting upon spine load predictions using a 2 muscle vs. 10 muscle EMG-assisted model.

The study consisted of a 2x2 design where 30 male and female subjects lifted 3 weights from 18 lift origins. Lifts were performed using both a one-handed and two-handed lifting (hands) technique. A well-established EMG-assisted personalized model that includes active and passive length-force relationships, curved torso muscle trajectories, muscle motion, and muscle gain calibration was employed to assess the impact of including 2 muscle (erector spinae only) vs. the 10 power producing muscles of the spine. Only the effects “hands” and muscles included within the model (muscles) are reported here.

Both the “hands” factor and the “muscles” factor were statistically significant as well as their interaction. In general, one handed lifts resulted in 9% lower peak compression at L3/L4 and 13% lower A/P shear compared to two handed lifts at L5/S1. Model performance was excellent for both models. However, the number of muscle included in the model resulted in dramatically different forces. The 10 muscle model resulted in a 21% increase (500N) in compression and 23% increases in lateral shear. The interaction was characterized by the 10 muscle model identifying more pronounced differences in loading.

The differences in model results were large enough to drive spine loads above tolerance limits for much of the population when the more complete 10 muscle model was used compared to the two muscle model. These results highlights the importance of including more complete (10 muscle) and personalized (EMG-assisted) models in estimating spine forces.

Estimating Lumbar Passive Stiffness Behaviour from Subject-Specific Finite Element Models and In Vivo 6DOF Kinematics

Affolter C, Kedzierska J, Vielma T, Aiyangar A

Mechanical Systems Engineering, EMPA-Swiss Federal Laboratories for Materials Science and Technology, Duebendorf, Switzerland

Passive rotational stiffness of the osseo-ligamentous spine is an important input parameter for estimating in-vivo spinal loading using musculoskeletal models. These data are typically acquired from cadaveric testing [1]. Increasingly, they are also estimated from subject-specific imaging-based finite element (FE) models, which are typically built from CT/MR data obtained in supine position and employ pure rotation kinematics [2, 3]. In the current study, we explored the sensitivity of FE-based lumbar passive rotational stiffness to two aspects of functional in-vivo kinematics: (a) accounting for passive strain changes from supine to upright standing position, and (b) in-vivo coupled translation-rotation kinematics.

We developed subject-specific FE models of four subjects' L45 functional spinal units from supine CT images. We then simulated sagittally symmetric flexion in two ways: (i) pure flexion up to 12° under a follower load of 500N directly from the supine pose. (ii) First, a displacement-based approach was implemented to attain the upright pose, as measured using Dynamic Stereo X-ray (DSX) imaging [4]. Following this, we simulated in-vivo flexion using DSX imaging-derived sagittally symmetric coupled rotation-translation kinematics. Datasets from weight-bearing motion with three different external weights [10 lb (4.5 kg), 20 lb (9.1 kg), 30 lb (13.6 kg)] were used as inputs. No external loading was applied to the FE models.

Accounting for motion (and accumulated pre-strain) from supine to upright standing generated compressive pre-loads ≈ 650N (±309N) (Fig. 1). Additionally, a rotational "pre-torque" ≈3.5Nm (±1.3Nm) torque, on average, was also generated, corresponding to 22.5% of the reaction moment generated at 10° of L45 flexion. Rotational stiffness estimates obtained from DSX-based coupled translation-rotation kinematics were substantially higher compared to pure flexion simulation results. Reaction Moments were 80% and 48% higher at 5° and 10° of L45 flexion respectively (Fig. 2). Within-subject differences in rotational stiffness based on external weight were small, although between-subject variations were large.

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[3] Schmidt, H., Heuer, F., Drumm, J., Klezl, Z., Claes, L., and Wilke, H.-J., 2007, "Application of a calibration method provides more realistic results for a finite element model of a lumbar spinal segment," Clinical biomechanics (Bristol, Avon), 22(4), pp. 377-384.

[4] Aiyangar, A. K., Zheng, L., Tashman, S., Anderst, W. J., and Zhang, X., 2014, "Capturing three-dimensional in vivo lumbar intervertebral joint kinematics using dynamic stereo-X-ray imaging," Journal of Biomechanical Engineering, 136(1).

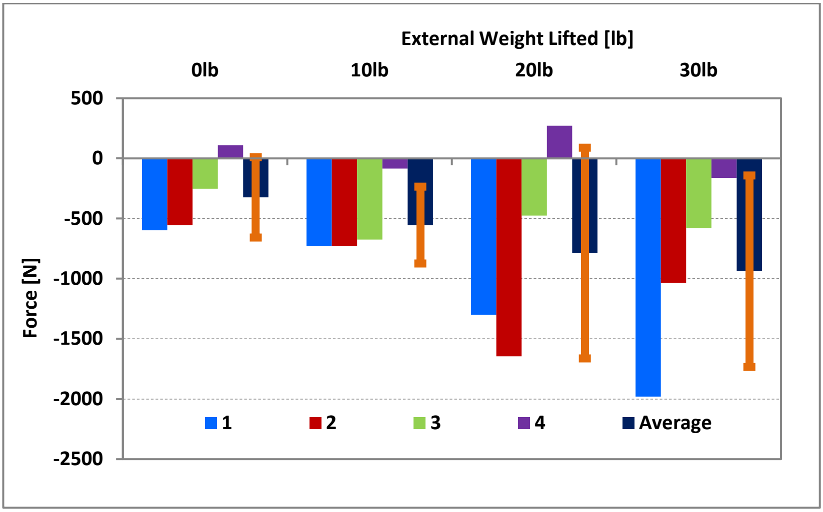

Fig. 1: Compressive pre-load generated between CT-derived supine pose and upright standing pose with external weight in hand.

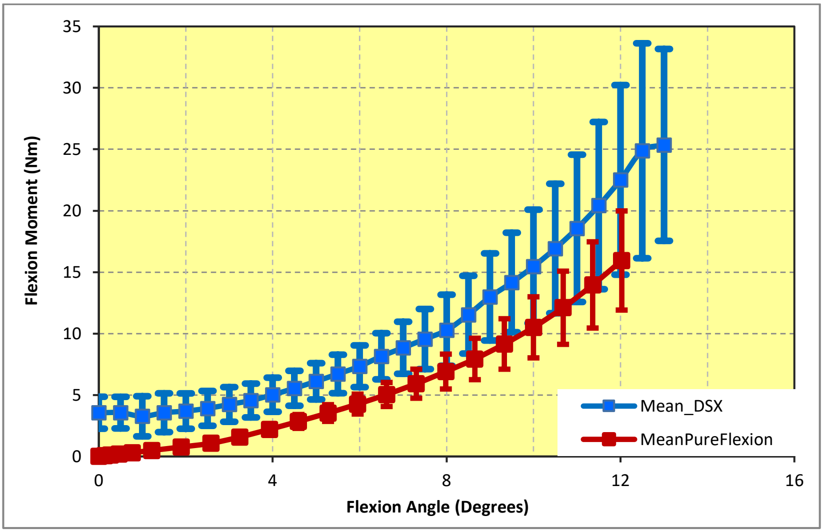

Fig. 2: FE-computed reaction moment-angular displacement (flexion) curves based on (a) Pure flexion rotation, and (b) DSX-based coupled rotation-translation kinematics.

A Novel Method for Prediction of Postoperative Global Sagittal Alignment based on Full-Body Musculoskeletal Modeling and Posture Optimization

Ignasiak D

NuVasive, Inc., San Diego, CA, USA

As sagittal imbalance is related to pain and disability, spinal fusion surgery aims to restore spinal alignment. The surgical outcomes might be critically affected by the reversal of compensatory changes, but its extent remains challenging to estimate preoperatively. The purpose of this study is to propose a method for predicting full-body sagittal alignment based on a treatment plan, using musculoskeletal modeling and inverse-inverse dynamics approach.

The pre- and postoperative data of adult spinal fusion patients were obtained retrospectively from an ongoing clinical study. An established full-body model (AnyBody) was used, with fused segments modeled as rigid. The generic model was modified to represent patient-specific sagittal spino-pelvic alignment, body weight and height, muscle deterioration, and details of the underwent corrective treatment (Fig. 1). Inverse-inverse dynamic simulations were performed to optimize the posture and predict reciprocal changes in unfused body parts. Based on the concept of the cone of economy, minimal muscle expenditure was used as posture optimality criterion. Predicted postural changes were compared to follow-up radiographs to evaluate method validity.

Twenty-one cases were analyzed in this preliminary study (age range = 48-74; number of fused segments 1-14). The model predictions correlated well with the radiographic measures at follow-up: T1-pelvic angle, TPA, $r = 0.83$; pelvic incidence – lumbar lordosis mismatch, ΔPILL, $r = 0.90$ (Fig. 2); lumbar lordosis, LL, $r = 0.90$; thoracic kyphosis, TK, $r = 0.77$. The model demonstrated high accuracy in predicting sagittal imbalance (positive predictive value = 1.00, negative predictive value = 0.75).

Study limitations include: small sample size, muscle expenditure definition, model assumptions and disregarding non-mechanical factors. Nevertheless, this is the first preoperative planning approach based on full-body biomechanical analysis. Following more extensive validation, this method could be applied to optimize patient-specific treatment plans, improving outcomes of spinal fusion surgery.

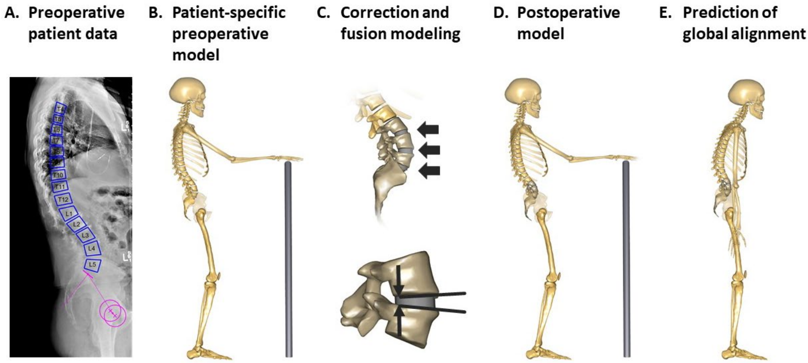

Fig. 1: Illustration of the modeling approach for prediction of postoperative global sagittal alignment. Preoperative patient data (A), including long lateral radiograph, body weight, height, BMI, sex, age and spinal pathology, are used to construct preoperative model (B), that reflects patient-specific spino-pelvic alignment, body size and muscle quality. (C) The correction at selected spinal levels is represented by adjusting intervertebral angles and heights, and fusion of these segments is modeled by introducing rigid constraints between vertebrae (marked with darker shade). In this way, patient- and treatment- specific postoperative model is constructed (D). This model is used to predict the global alignment changes due to introduced correction, including reciprocal changes at untreated lumbar, thoracic and cervical segments, pelvis and lower limbs (E). This is achieved by performing inverse-inverse dynamics simulations optimizing the posture based on muscle expenditure minimization.

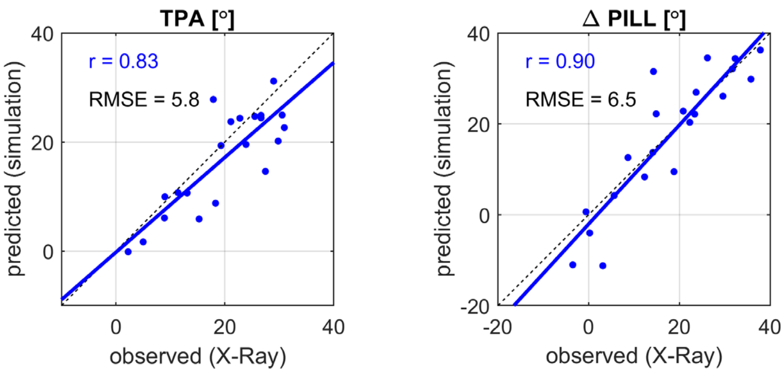

Fig. 2: Comparison between postural measures predicted by the simulation and observed in follow-up radiographs. Pearson's correlation coefficients (r) and root-mean-square-error (RMSE) values are displayed.

Predicting Intervertebral Disc Loading and Trunk Muscle Activity in Healthy Adolescents using Musculoskeletal Full-Body Models

Schmid S[a,b,c], Allaire BT[a], Burkhart KA[a,d], Grindle D[a], Bouxsein ML[a,b], Anderson DE[a,b]

[a]*Beth Israel Deaconess Medical Center, Center for Advanced Orthopaedic Studies, Boston, MA, USA*
[b]*Harvard Medical School, Dept. of Orthopaedic Surgery, Boston, MA, USA*
[c]*Bern University of Applied Sciences, Dept. of Health Professions, Bern, Switzerland*
[d]*Massachusetts Institute of Technology, Harvard-MIT Health Sciences and Technology Program, Cambridge, MA, USA*

Currently available thoracolumbar spine models are entirely based on data from adults and might therefore not be applicable for simulations in adolescents. We therefore created and validated musculoskeletal full-body models including a detailed thoracolumbar spine for adolescents and validated by comparing segmental loading and muscle activity predictions to in vivo data.

Our recently developed adult thoracolumbar spine model was combined with a lower extremity model (Fig. 1). Adolescent models were created for each year from 10-18 years of age by adjusting segmental length and mass distribution, center of mass positions and moments of inertia of the major body segments as well as sagittal pelvis and spine alignment based on literature data. Similarly, muscle strength properties were adjusted based on CT-derived cross-sectional area measurements.

Simulations were conducted from two in vivo studies: 1) an 11-year-old model to predict disc height change (lumbar disc compressibility, LDC) when carrying different backpack loads [1] and 2) an 18-year-old model to predict L3/4 intradiscal pressure (IDP) and trunk muscle activity in different body positions [2]. Correlations between model predictions and in vivo data from the literature were analyzed using linear regression models.

Predicted LDC values correlated well with in vivo data for all lumbar levels (R^2>0.84) with a tendency for underestimation, except from the T12/L1 level (Fig. 1). Predicted values also correlated well for IDP (R^2=0.87) and muscle activity, particularly for erector spinae (R^2=0.81) (Fig. 2).

The results indicate the suitability of our models for the reasonably accurate prediction of segmental loading and trunk muscle activity in healthy adolescents. When aiming at investigating adolescent populations with pathologies such as idiopathic scoliosis, our models can serve as a basis for the creation of deformed spine models as well as for comparative purposes.

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[2] Schultz et al. *J Bone Joint Surg Am*. 1982;64:713-20.

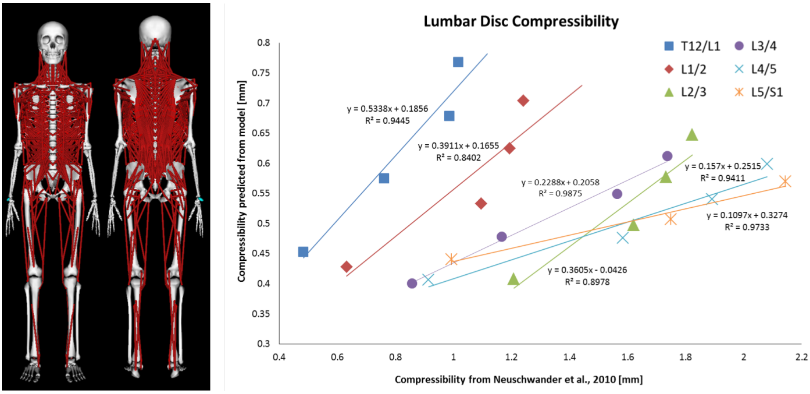

Fig. 1: Left: Musculoskeletal full-body model including a detailed thoracolumbar spine. Right: Disc compressibility on the T12/L1, L1/2, L2/3, L3/4, L4/5 and L5/S1 levels when standing upright and carrying a backpack with different loads (4, 8 and 12kg). In vivo data were retrieved from Neuschwander et al. [1] for adolescents (mean age: 11 years) undergoing upright standing MRI analyses.

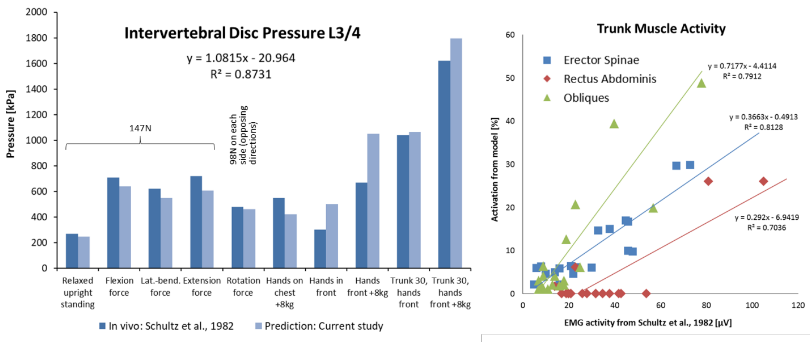

Fig. 2: Intradiscal pressure at the L3/4 level and trunk muscle activity when resisting forces on the T6-level and in different upper body positions. In vivo data were retrieved from Schultz et al. [2] for young adults (mean age: 21.8 years) having a transducer inserted in the third lumbar disc.

Coupled Artificial Neural Networks to Predict Whole Body Posture, Lumbosacral Moments, Trunk Muscle Forces, and Lumbar Disc Loads during Three-dimensional Material Handling Activities

Aghazadeh F, Arjmand N, Mohammadi A, Soufi AM

Dept. of Mechanical Engineering, Sharif University of Technology, Tehran, Iran

Manual material handling (MMH) is associated with mechanical back injures. To manage such injuries, musculoskeletal models are employed to estimate spinal loads during MMH. These models require, as input, 3D-position of the hand-load and body posture whose measurements involve skillful time-consuming motion analysis investigations.

To facilitate the procedure of posture measurements and load estimations, three coupled artificial-neural-networks (ANNs) were developed. To predict whole body posture, ANN_1 was trained based on our novel measurements on 15 individuals. Each individual performed 135 static-tasks by holding 0, 5, and 10 kg weight at 9 different anterior-left positions and at 5 heights (0, 30, 60, 90, and 120 cm/floor). Posture was measured via the Simi Reality Motion Systems that recorded 3D-position of 15 skin-markers on the head, C7, T12, L5, pelvis, left/right shoulders, elbows, wrists, knees and ankles. The ANN_1 predicted posture by identifying the relationship between 5 inputs (hand-load magnitude, its 3D-position and body height) and 45 outputs (3D-position of markers); total of 91125 input/output datasets: 15 subjects$\times$135 tasks$\times$45 marker-positions. Moreover, for each subject/task, the 3D L5-S1 external moments were evaluated using the measured posture, hand-load position/magnitude, and anthropometric data [1]. The ANN_2 identified the relationship between the foregoing inputs (plus body weight) and L5-S1 moments; total of 6075 datasets: 15 subjects$\times$ 135 tasks $\times$3 moments. Finally, predicted posture by ANN_1 and hand-load position/magnitude were input into a previously-developed/validated ANN_3 [2] that predicted lumbar disc loads and trunk muscle forces (Fig. 1).

Trained ANNs could predict 3D body posture (R^2= 0.974/RMSE= 7 cm) and L5-S1 moments (R^2= 0.972/RMSE= 16.6 Nm) from novel sets of inputs that were not included in the training processes (Fig. 2).

For any given MMH task, the coupled ANNs were capable of predicting body posture, L5-S1 moments, muscle forces and lumbar loads via easy-to-measure inputs.

[1] Plamondon et al., 1995. Clin Biomech 10:128-36.

[2] Arjmand et al., 2013. J Biomech. 46:1454-62.

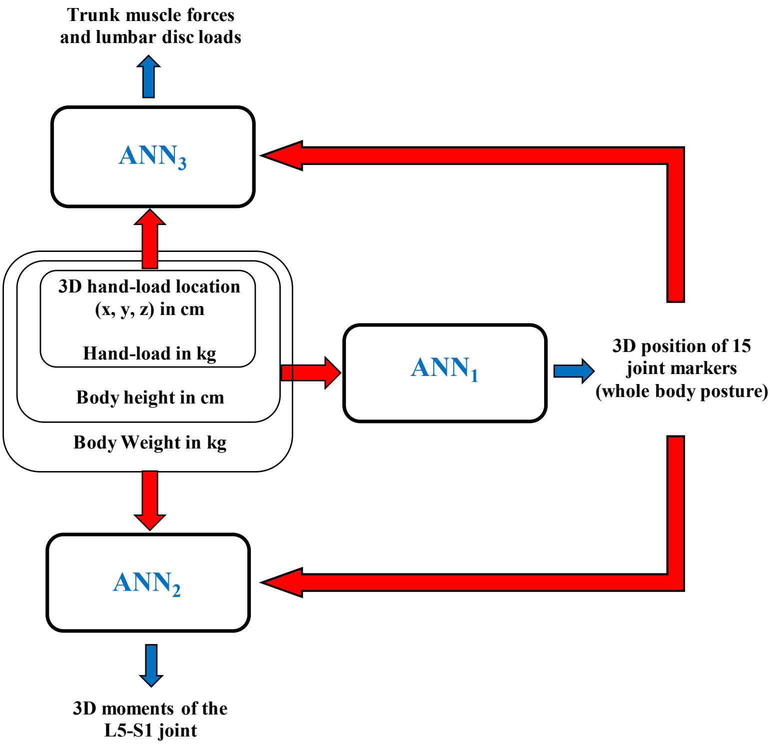

Fig. 1: A schematic of the coupled ANNs.

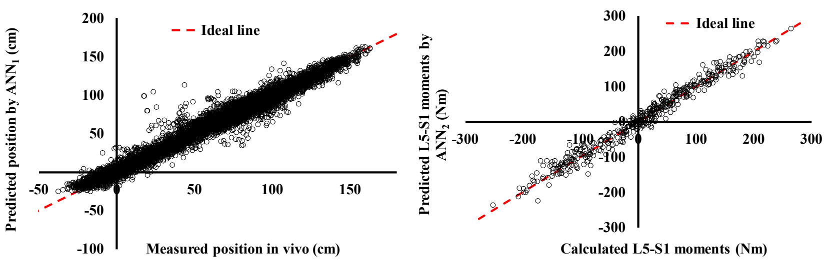

Fig. 2: Scatter plots of all target values (horizontal axis) and their corresponding outputs of the (a) trained ANN_1 for body posture prediction and (b) trained ANN_2 for the L5-S1 moment prediction from novel sets of inputs that were not included in the training processes showing the ANN's capability to generalize.

Influence of Seat Parameters on Computationally Predicted Spine Loading

Theodorakos I, Beurier G, Wang X

Université Lyon, IFSTTAR, LBMC UMR_T 9406, Lyon, France

Andersson et al. [2, 3, 4] reported decreased experimentally measured L3L4 intra-disc pressure, as the backrest was backwards inclined. On the other hand, minor differences were reported on the L3L4 intra-disc pressure for different seat pan angles. Although such experimental studies provide great insight into how spine loading is influenced by different seat parameters, they are invasive and expensive. Alternatively computational models can be employed to provide spine loading estimations in a cheap and fast manner. Rasmussen et al. [7] and Grujicic et al. [5] employed a musculoskeletal (MSK) model to investigate the influence of seat pan/buttocks friction coefficient, the backrest and the seat pan inclination on spinal loading. These computational studies reported complex effects among the tested variables on the spinal loading. The present study investigated the influence of seat parameters on spinal loading, employing a personalized MSK model.

Experimentally obtained kinematic and force data [8] served as input for a MSK model (Fig. 1) that enabled the computation of spinal reaction forces. Statistical analysis was performed to investigate the influence of the seat pan angle (0°, 5° preferred) and backrest angle (100°, 110°) on computationally predicted spinal forces. A significant reduction on the L4L5 compression force was observed for the 110° backrest angle compared to the 100°. This can be explained from the fact that the more inclined the backrest is the more of the body weight is supported by the backrest, resulting in less spinal loading. Moreover, significantly increased L5S1 compressive force was observed for the 0° seat pan angle compared to 5° and the preferred seat pan angles, indicating that participants selected a seat pan angle that reduced the spinal load. Our results demonstrate the potential of computationally predicted loads to provide insight into the interaction of the human body with its environment and to be employed for seat designing.

[1] Andersen, M.S., Damsgaard, M., MacWilliams, B., Rasmussen, J., 2010. A computationally efficient optimisation-based method for parameter identification of kinematically determinate and over-determinate biomechanical systems. Comput Methods Biomech Biomed Engin 13, 171-183.
[2] Andersson BJG, Ortengren R, Nachemson A, Elfstrom G., 1974a. Lumbar disc pressure and myoelectric back muscle activity during sitting. I. Studies on an experimental chair. Scand J Rehabil Med; 6:104-14.
[3] Andersson BJG, Ortengren R., 1974b. Lumbar disc pressure and myoelectric back muscle activity during sitting. III. Studies on a wheel chair. Scand J Rehabil Med; 6:122-7.
[4] Andersson BJG, Ortengren R, Nachemson A, Elfstrom G., 1974c. Lumbar disc pressure and myoelectric back muscle activity during sitting. IV. Studies on a car driver's seat. Scand J Rehabil Med; 6:128-33.
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[6] Nerot, A., Skalli, W., Wang X., 2016. A principal component analysis of the relationship between the external body shape and internal skeleton for the upper body. J. Biomech 49 (14), 3415–3422.
[7] Rasmussen, J., Torholm, S., de Zee, M., 2009. Computational analysis of the influence of seat pan inclination and friction on muscle activity and spinal joint forces. Int J Ind Ergon 39, 52–57.
[8] Wang, X., Cardoso, M., Beurier, G., 2018. Effects of seat parameters and sitters' anthropometric dimensions on seat profile and optimal compressed seat pan surface. Appl` Ergon 73, 13-21.

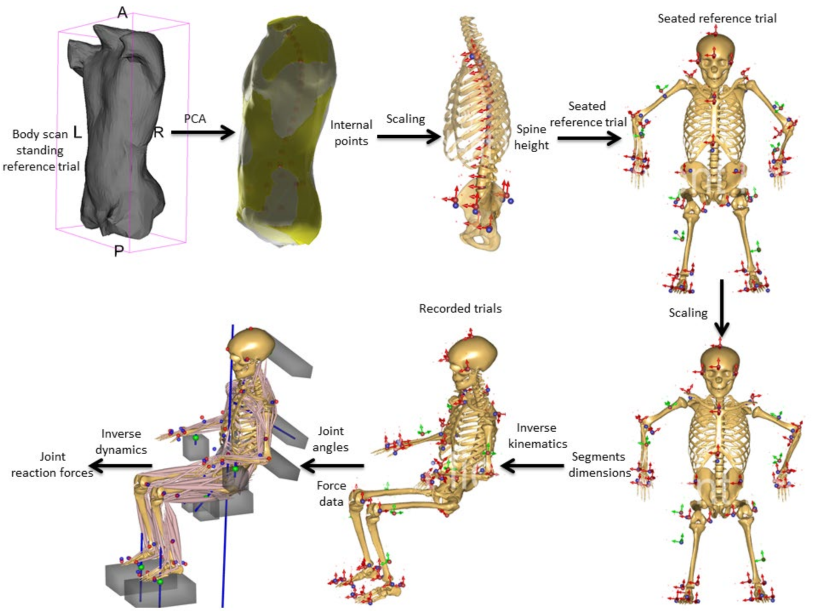

Fig. 1: Modeling procedures followed to obtain spine reaction forces from experimentally obtained data. Initially the spine was personalized using the participants' external body shape recorded in a body scanner [6]. The rest of the segments were personalized based on a seated reference trial, using an optimization procedure [1]. Then, the joint angles were obtained by an inverse kinematics analysis. The computed joint angles and the measured contact forces were served as inputs in an inverse dynamics analysis that allowed the computation of the joint reaction forces.

Statistical Shape Model Predicted Alignments and Musculoskeletal Simulation in Surgical Planning

Senteler M[1,2], Caprara S[1,2], Gschwend O[1], Snedeker JG[1,2], Farshad M[1]

[a]*Dept. of Orthopaedics, Balgrist Hospital, University of Zurich, Zurich, Switzerland*
[b]*Institute for Biomechanics, Swiss Federal Institute of Technology (ETH), Zurich, Switzerland*

Patient specific musculoskeletal models proofed to be a powerful tool for assessing spinal loading and evaluate surgical treatment strategies [1]. During surgical planning, however, the ideal alignment is often not known and choosing the amount of correction is mostly based on surgeon's experience. Furthermore, it remains unclear how the loading at spinal segments is affected by the intervention, although potentially clinically relevant.

We aimed at predicting postoperative alignment using a statistical shape model (SSM) based method, hypothesizing that SSM predicted alignment leads to favorable loading conditions in the lumbar spine.

A statistical shape model for predicting sagittal alignment of the spine based on the position of femoral heads (FH) and the sacrum was created and trained with data from a set of 50 lateral EOS images. Annotations included FH, sacral superior endplate, and endplates for each vertebral body from C1 to L5. For each spinal level the vertebral center point was calculated and served as training data for the SSM.

The trained SSM was employed to predict a sagittal alignment for a low back pain patient undergoing spinal fusion surgery. A patient specific musculoskeletal model was created in OpenSim for the insitu and the SSM-predicted alignment by adjusting a validated template model [2]. Joint reaction analysis was performed for linear forward bending motion from upright standing to 30° lumbar flexion (50° upper body flexion).

The SSM-predicted alignment features considerably larger lumbar lordosis and thoracic kyphosis (Fig. 1), and re-established sagittal balance as determined by sagittal vertical axis (C7 plumb line on S1).

Simulated compression forces for the SSM fitted alignment were lower in all postures at all levels (Fig. 2), except for level L12 in upright standing configuration, which caused similar forces as the original alignment (306 N vs. 305 N, respectively).

At levels L12 to L45 the magnitude of experienced shear forces throughout all lumbar levels for the SSM predicted alignment was within the range of those predicted for the original alignment; only at level L5S1 in flexed postures the shear force was considerably higher in the SSM alignment.

The present study demonstrated the ability of SSM models to define spinal alignments that reduce the loading at lumbar segments. The framework proofed its potential in supporting surgical planning, which is currently addressed in a retrospective analysis.

[1] Senteler et al. 2017, Fusion angle affects intervertebral adjacent spinal segment joint forces, J. Orthop. Res. 35, 131–139
[2] Senteler et al. 2016, Intervertebral reaction force prediction using an enhanced assembly of OpenSim models. Comput. Methods Biomech. Biomed. Engin. 19, 538–548.

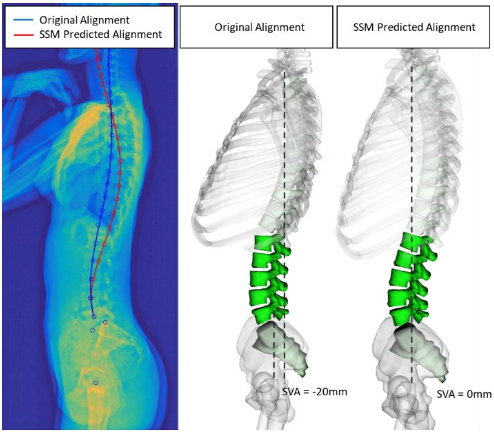

Fig. 1: EOS with original and SSM predicted alignment (left) and resulting alignments in musculoskeletal models.

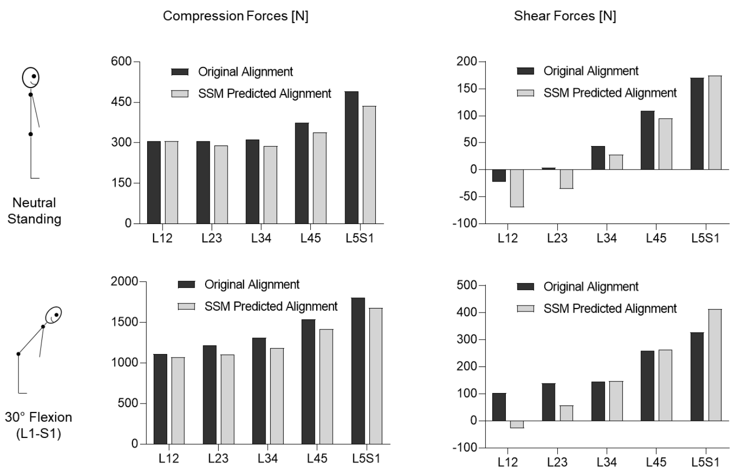

Fig. 2: Compression (left) and shear forces (right) for as predicted by a patient specific musculoskeletal model featuring original and SSM predicted sagittal spinal alignment. Simulation covered linear movement from upright to 30° lumbar flexion, results are shown for the initial (top) and the final posture (bottom).

Trunk Stabilization in Patients with Low-back Pain and Healthy Controls

van Dieën JH and Griffioen M

Dept. of Human Movement Sciences, Vrije Universiteit Amsterdam, Amsterdam Movement Sciences, Amsterdam, The Netherlands

While inadequate stabilization of the trunk is often invoked as a potential cause of low-back pain and is a target of many exercise approaches used in low-back pain management, evidence on inadequate quality of trunk control in low-back pain patients is inconsistent. The aim of this project was to compare trunk stabilization between patients with low-back pain and healthy controls. To this end, we have developed a systems identification method to assess trunk stabilization, which is based on robot-controlled unpredictable, force-controlled perturbations of upright sitting trunk posture as an input and trunk movement and trunk extensor EMG activity as outputs. We used this method to assess trunk stabilization in 50 healthy subjects and 49 patients with low-back pain. Experiments were performed under two task instructions: to maximally resist the perturbations and to relax but remain upright. Results indicated higher resistance to perturbations or lower admittance in low-back patients than in healthy controls, especially when asked to relax. The lower admittance in low-back pain patients was mainly caused by higher reflex gains. The results suggest that, in contrast to common beliefs, patients are capable of stabilizing upper body posture at least as good as healthy subjects, but that they control trunk posture more rigidly when asked to relax. This was corroborated by the fact that patients showed less modulation of reflex gains between tests with different instructions. In low-back pain patients, admittance and reflex gains were correlated with pain-related beliefs and may hence be mediated by psychological pain responses. In conclusion, low-back pain patients appear to maximize control over trunk posture, at the cost increased control effort and spinal loading.

Indication of Diagnostic Criteria for Proprioception Disorders between Non-Specific Low Back Pain Patients and Healthy People based on Analysis of Linear and Nonlinear Parameters of Center of Pressure and Trunk Stability

Shokouhyan SM[a], Davoudi M[a], Talab MH[a], Abedi M[b], Bervis S[c], Parnianpour M[a]

[a]*Dept. of Mechanical Eng, Sharif University of Technology, Tehran, Iran*
[b]*Dept. of Rehabilitation, Shahid Beheshti University, Tehran, Iran*
[c]*Dept. of Rehabilitation, Shiraz University of Medical Sciences, Shiraz, Iran*

Postural stability is one of the health indicators of musculoskeletal system. CNS uses vision, vestibular and somatosensory information to maintain body stability. Researches have shown that LBP patients use more ankle proprioception than lumbar proprioception which explains less reliance on lumbar proprioception.

In this study, in order to achieve the aim of altering proprioception, vibrators with frequency of 70 Hz and amplitude of 0.5 millimeter were used. This study was performed on two groups of 20 healthy people and non-specific low back pain patients. A vibrator was placed on the Soleus muscle area of each legs and two vibratos were also placed on the area of lower back muscles bilaterally. Individuals were placed on surface of force plate and trunk angles were also recorded simultaneously. Tests were performed in 8 trials which independent variables were vibration (4 levels) and surface (2 levels: foam and rigid) for within subjects and 2 groups (healthy and LBP patients) for between subjects (4×2×2). Subject's vision was occluded during the trials. Linear COP parameters (deviation of amplitude, deviation of velocity, Phase plane portrait, and overall mean velocity) and nonlinear parameters (RQA and Lyapunov exponents) were chosen as dependent variables. All dependent variables were subject to multi factor ANOVA and subsequent Bonferroni adjusted multiple comparison of levels of independent variables.

Results were shown that linear parameters were higher in patients than healthy subjects, indicating that they were more willing to use ankle strategy, which was consistent with previous researches. RQA parameters for the center of pressure on both sides and for the trunk sagittal angle (Determinism and Entropy) indicated more repeated patterns of movement by the patient, suggesting that these individuals, insist on using a repeat strategy (Ankle) even the surface is unstable (foam). Analysis of short and long Lyapunov exponents for the center of pressure and trunk angle showed that people with low back pain caused no use of all joints in their bodies (Non-flexible), are less stable than healthy subjects. Statistical analysis showed a significant difference between linear parameters and RQA and Lyapunov exponents between healthy and LBP patient. These criteria could be used to diagnose those with proprioception problems and to observe outcomes patient's treatment process during rehabilitation levels.

Sudden Gait Perturbations elicit Sex-specific Neuromuscular Trunk Responses in Persons with Low Back Pain

Mueller J[a], Martinez-Valdes E[b], Mueller S[a], Kulig K[c], Mayer F[d]

[a]*Exercise Science and Applied Biomechanics, Dept. of Computer Science – Therapy Science, Trier University of Applied Science, Trier, Germany*

[b]*School of Sport, Exercise and Rehabilitation Sciences, Centre of Precision Rehabilitation for Spinal Pain (CPR Spine), College of Life and Environmental Sciences, University of Birmingham, Birmingham, UK*

[c]*Division of Biokinesiology and Physical Therapy, University of Southern California, Los Angeles, CA, USA*

[d]*University Outpatient Clinic, Sports Medicine & Sports Orthopaedics, University of Potsdam, Potsdam, Germany*

Persons with Low Back Pain (LBP) exhibit delayed trunk muscle onset and increased co-contractions as a response to quasi-static and dynamic sudden trunk loading in comparison to back-healthy controls (BHC). Although LBP is more prevalent in females, sex-specify responses have not been well documented. Therefore, the purpose of the study was to explore sex-specific neuromuscular differences, to gait perturbation, in persons with LBP.

Twenty-nine LBP subjects (12m/17f;31±10yrs;174±12cm;71±16kg) walked on a split-belt treadmill at 1m/s, while 15 right-sided random perturbations (treadmill belt decelerating, 40m/s^2, 50ms duration; 200ms after heel contact) were applied. Trunk muscle activity was assessed using a 12-lead surface EMG (6 back and 6 abdominal muscles; 4000Hz). EMG-RMS [%] (0-200ms after perturbation) was calculated and normalized to RMS of unperturbed gait for each muscle. Furthermore, muscle onsets (T; ms) were determined. Two-way ANOVA (factors: sex/ muscle) was applied to account for sex differences in main outcomes.

EMG-RMS (amplitudes; mean) ranged from 356% to 901% in males and 349% to 694% in females representing a significant interaction effect (sex*muscle: p=0.017). Post-hoc analysis revealed significant differences for EMG-RMS analysis of rectus abdominis right (p=0.043; f>m) as well as obliques externus right/left (p=0.018/p=0.005; f<m). In the time domain, females show shorter (mean: 90±16ms) response times (T) compared to males (mean: 98±22ms) in all 12 trunk muscles without significant interaction effect (sex*muscle: p=0.9).

In this LBP population, abdominal muscle activation discriminated females from males. Specifically, females had higher activity of the rectus abdominis and lower activation of the oblique muscles. These different activation strategies might be relevant to the development of sex-specific intervention strategies.

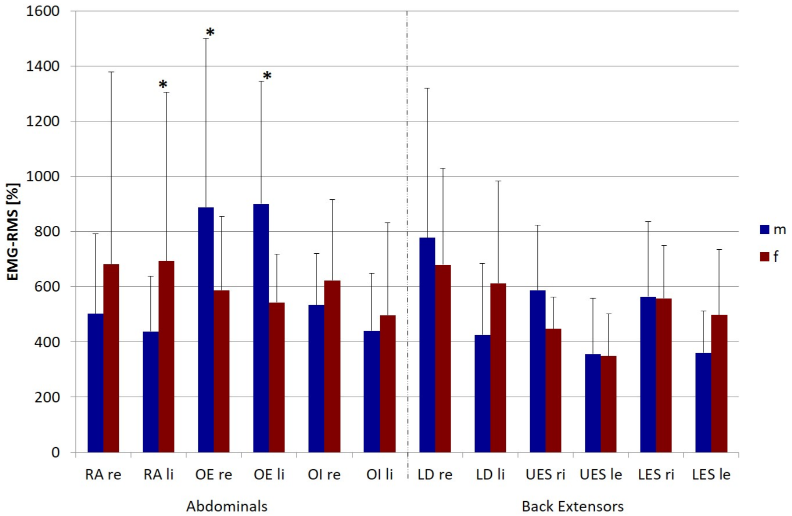

Fig. 1: EMG RMS [%] 0-200ms after perturbation normalized to normal gait for all trunk muscles (mean±SD) in male and female LBP patients.
Legend: Trunk muscles: Mm rec. abd. (RA), obl. ext. abd. (EO), obl. int. abd (IO) of left and right side; erec. spinae thoracic (T9; UES)/ lumbar (L3; LES), latis. dorsi (LD); Le = left side; ri = right side; * significant sex differences (p<0.05).

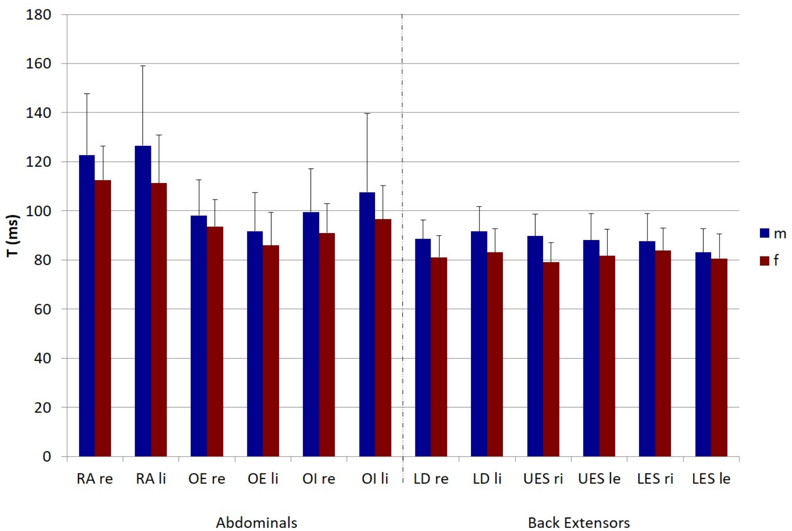

Fig. 2: Muscle onset (T; ms) for all 12 trunk muscles in male and female LBP patients.
Legend: Trunk muscles: Mm rec. abd. (RA), obl. ext. abd. (EO), obl. int. abd (IO) of left and right side; erec. spinae thoracic (T9; UES)/ lumbar (L3; LES), latis. dorsi (LD); le = left side; ri = right side.

Can Trunk Postural Control During Unstable Sitting be considered a Proxy Measure of Dynamic Lumbar Stability?

Larivière C[a,b], Preuss R[b,c], Henry SM[d]

[a]*Institut de recherche Robert-Sauvé en santé et en sécurité du travail (IRSST), Montréal, Québec, Canada*

[b]*Center for Interdisciplinary Research in Rehabilitation of Greater Montréal (CRIR), Montréal, Québec, Canada*

[c]*School of Physical & Occupational Therapy, McGill University, Montréal, Québec, Canada*

[d]*Dept. of Rehabilitation Therapy, The University of Vermont Medical Center, Burlington, VT, USA*

Measuring dynamic lumbar stability remains elusive, constraining researchers to measure its determinants individually (Fig. 1). Trunk postural control during unstable sitting has many of the same determinants as dynamic lumbar stability, and may be an appropriate proxy measure. We aimed to test this hypothesis.

Subjects (n = 64) sat on a wobbling chair for 60-s trials. Chair motion was quantified with an inertial sensor, and six outcomes of sway performance were computed (Table). Variables extracted from five additional tests were linked with lumbar stability determinants (Fig. 1: 1) lumbar proprioception during trunk axial rotation; 2) lumbar intrinsic and reflective stiffness during trunk perturbations; 3) ultrasonographic thickness measures of back (LuM, thoracolumbar fascia) and abdominal (EO, IO, TrA and fascia separating them) structures at rest, and muscle thickness change during contraction; 4) EMG onsets of trunk muscles (feedforward control) and their mechanical effects (trunk kinematics) following rapid arm movements; and 5) flexion-relaxation phenomenon and coordination of trunk segments (pelvis, lumbar and thoracic) during maximal trunk flexion. The six sway performance outcomes were each regressed, using a forward stepwise procedure, with the test variables as candidate predictors.

Across the six performance outcomes, 4 or 5 predictors explained between 36 and 47% of outcome variance. Predictors were associated with factors F1 to F5 of 12 factors defined by PCA (see Fig. 1), as well as the percent change of TrA thickness.

The ability of lumbar stability determinants to explain a high proportion of variance in measures of sitting balance supports the use of sitting balance as a proxy measure of dynamic lumbar stability. Only variables of structural function (F1, F2, F3, F5: related to intrinsic stiffness) and of muscle activation and coordination (F4: APA, F3: reflexive stiffness) were predictive. No variable of transducer function (e.g., proprioception) was predictive.

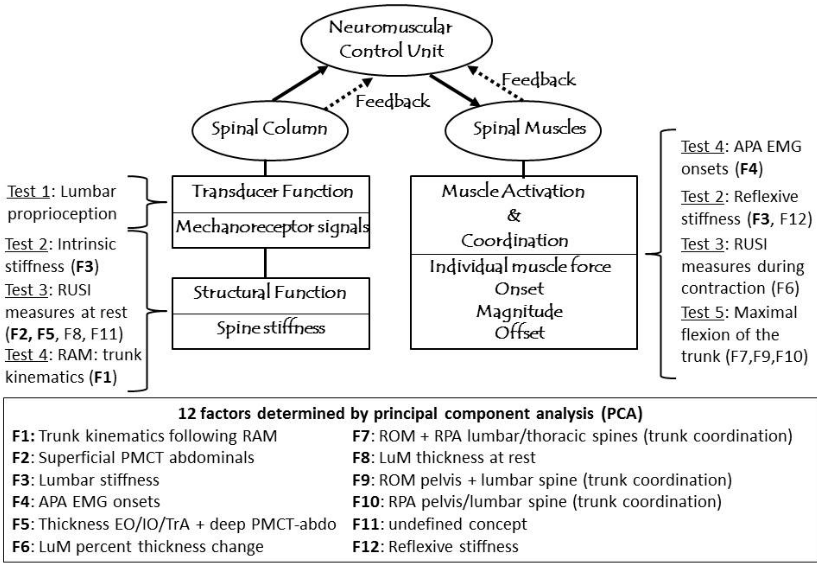

Fig. 1: Upper center drawing: Determinants of spine stability as theoretically modeled [Panjabi (2006). *Eur. Spine J.* 15: 668-676]. Left and right text columns: The 5 neuromuscular tests carried out to measure these determinants. Text box at the bottom: The 12 factors (and labelling) corresponding to the grouping of the numerous variables (PCA) extracted from these tests. **APA**: anticipatory postural adjustments; **EO** and **IO**: external and internal obliques; **LuM**: lumbar multifidus; **PMCT-abdo**: perimuscular connective tissues around abdominals; **RAM**: rapid arm movement; **ROM**: range of motion; **RPA**: relative phase angle; **RUSI**: rehabilitative ultrasound imaging; **TrA**: transversus abdominis.

Abbreviation	Description
Prieto et al. (1996). *IEEE Trans.Biomed.Eng.*, 43:956-966.	
MFREQ (Hz)*	Mean frequency
MVELO (°/s)*	Mean velocity
FREQD *	Frequency dispersion
Sway density analysis: Baratto et al. (2002). *Motor Control.*, 6:246-27	
meanDUR (s)	Mean time between consecutive SDC peaks
meanDIST (°)	Mean of the spatial distance between consecutive SDC peaks
Rosenstein et al. (1993). *Physica D: Nonlinear Phenomena.*, 65:117-134.	
LyapunovS	Lyapunov exponent corresponding to the short interval

* These measures are computed based on the radial distance (RD) time series (i.e., the vector distance from the mean COP to each pair of points in the AP and ML time series).
SDC: Sway density curve

Table: Six selected performance outcomes of sitting balance.

Biomechanics of Intra-Abdominal Pressure in Spine Stiffening and Loading - A Systematic Review of In-Vivo and Modeling Studies

Arjmand N, Dehghan-Hamani I, Khoddam Khorasani P

Dept. of Mechanical Engineering, Sharif University of Technology, Tehran, Iran

Activation of trunk muscles during physical activities pressurizes the intra-abdominal cavity thereby elevating the intra-abdominal pressure (IAP). An elevated IAP is suggested to both stiffen and unload the spine. Mechanism and effectiveness of IAP in stiffening and unloading the spine remain, however, controversial.

A systematic review was performed. Keywords “IAP” and (“spine” or “trunk” or “muscle” or “stability” or “model” or “EMG”) were used to search PubMed databases since 1980. Inclusion criteria was studies on the spine/trunk of healthy humans. All studies on patients/animals, lower limb/pelvis/renal/urinary/respiratory systems, and measurement techniques were excluded. Reference lists of the final-included articles were screened (manual search) for studies that may have been missed by the electronic search.

The electronic search retrieved 1289 articles. After reading titles and abstracts, while considering inclusion/exclusion criteria, 77 articles were included for full-text screening. After the full-text screening and manual search, 62 articles were considered for this review study. This included 30 *in vivo*, 22 modeling, and 10 combined *in vivo*/modeling investigations. Preliminary results suggest that both *in vivo* and modeling studies generally confirm an increase in trunk stiffness as IAP elevates. IAP increases to stiffen the spine in a preparatory (i.e., prior to an exertion to help overcome inertia of rest) and/or a stability (i.e., during an exertion to help overcome mechanical perturbations) mode. Important controversies, however, exist both between and within modeling and *in vivo* studies as to the unloading mechanism of IAP. While some studies suggest that IAP unloads the spine by producing an upward force on the diaphragm and thus generating an auxiliary extensor moment, others indicate that abdominal co-activities, required to elevate IAP, produce a flexor moment that counterbalances the IAP-generated extensor moment.

Analysis of results indicates that effectiveness of IAP in stiffening and unloading the spine is task/posture specific.

Real-time Feedback to Reduce Lower Back Moment while Lifting a Box: a Proof-Of-Concept-Study

Punt M[a,b] , Nematimoez M[a,c], van Dieën JH[a], Bruijn SM[a], Kingma I[a]

[a]*Dept. of Human Movement Sciences, Vrije Universiteit, Amsterdam, The Netherlands*
[b]*Utrecht University of Applied Sciences, Utrecht, The Netherlands*
[c]*Dept. of Sport Biomechanics, Kharazmi University, Tehran, Iran*

Low back pain (LBP) is a very common health problem. Literature shows a clear exposure-response relation between work related lifting and LBP. In addition, it seems that a longer duration of lifting is associated with a higher LBP incidence. Trying to reduce low back loading may be an effective approach in preventing or reducing low back pain. Therefore this study investigated if real-time feedback on L5S1 moments during lifting reduces these moment while lifting and lowering a box.

We recruited 16 young healthy male participants without a recent history of low back pain and without prior biomechanical knowledge about lifting. Each participant performed 4 trials, each containing 12 lifting and 12 lowering tasks. Trial 1 was used as a reference trial to determine a threshold value, which was set at 80% of the average individual peak moment. Participants were unaware of what causes the feedback but were instructed to try to avoid the audio feedback sound by changing their lifting strategy in trials 2 and 3. Trial 4 was a retention test without feedback.

For both lifting and lowering, the peak moment at L5S1 and the time exceeding the threshold significantly reduced over trials (table 1). Furthermore, we found that lumbar flexion and trunk inclination significantly reduced over trials and knee angle significantly increased over trials.

Real time feedback on L5S1 moments is a promising approach to reduce low back loading during lifting and lowering. However, future work should examine whether similar results can be found in people with LBP.

Table 1

	Trial 1	**Trial 2**	**Trial 3**	**Trial 4**	**Trial**	**Position**	**Inter-action**
Lifting	Mean ± SD	Mean ± SD	Mean ± SD	Mean ± SD	P-value	P-value	P-value
Peak moment (Nm)	253.2 ± 57.7	239.3 ± 57.2	243.5 ± 61.7	230.5 ± 64.5	**<.001**	**.02**	.129
Time above threshold (ms)	266 ± 8.9	182 ± 8.5	225 ± 9.9	170 ± 10.9	**<.001**	**.004**	.494
Lowering							
Peak moment (Nm)	237.6 ± 53.2	225.9 ± 53.8	230.1 ± 58.6	217.1 ± 56.3	**<.001**	**.017**	.059
Time above threshold (ms)	365 ± 15	292 ± 15.3	309 ± 15.4	229 ± 15.9	**<.001**	**.001**	.321

Acknowledgements: This work was supported by the European Union’s Horizon 2020 through the SPEXOR project, contract no. 687662.

Reducing the Number of Input Variables Required to Control an Active Exoskeleton

Tabasi SA[a], Kingma I[a], de Looze MP[a,b], van Dijk W[b], Koopman AS[a], van Dieën JH[a]

[a]*Dept. of Human Movements Sciences, Vrije Universiteit Amsterdam, The Netherlands*

[b]*TNO, Leiden, The Netherlands*

Despite the development and use of modern technology, many jobs still require manual handling tasks. This is one of the causes that more than 40% of workers in the European Union continue to suffer from musculoskeletal pain [1]. A strategy to deal with work-related musculoskeletal disorders is the use of passive or actuated exoskeletons. For actuated exoskeletons, several control systems have been proposed based on different sets of input variables (e.g. kinematic or EMG data) [2]. One of the main challenges here is the feasibility of recording the required input variables in the occupational environment. The aim of this research is to find a selection of input variables to define the control strategy based on them while guaranteeing that the exoskeleton control is based on acurate estimation of biomechanical loads.

Eleven healthy participants lifted a 15kg box from mid-shin height, with various techniques (Stoop, Squat and Free), each of which was performed in four conditions, one without a device, and three with an actuated back-support exoskeleton (Robomate [3]) controlled by three different control strategies (trunk inclination, forearm EMG and a combination of these) and each condition repeated three times. During each trial, kinematics and EMG data of back and abdominal muscles were recorded and using inverse dynamics and an EMG-driven trunk muscle model, the active moment generated by the back muscles, which is the desired support by the exoskeleton, was calculated. To find a relation between input variables and the moment, random forest regression analysis was conducted for each subject individually. To train the model, the data of two repetitions of each condition were used and data of the other repetition were used to determine the quality of the model. After creating the model based on all the input variables, the importance of each was investigated and a regression model based on a reduced set of the most important variables was defined.

A regression model based on the longissimus thoracis muscle EMG signal, lumbar flexion angle, lumbar flexion velocity and trunk inclination angle could estimate active moments during all trials with $R2 >= 0.90$ for all the subjects but one ($R2 = 0.88$). This shows that having one EMG signal and three kinematic inputs suffices to estimate the support that should be provided by the exoskeleton.

In conclusion, it is possible to accurately estimate the active muscle moment, and thereby the desired assistive moment generated by the exoskeleton, based on a limited number of sensors. Further investigation is needed to find the minimum requirements on the training set needed for robust estimates.

[1] Eurofound, “Fifth European Working Conditions Survey, Publications Office of the European Union, Luxembourg.,” 2012.

[2] M. P. De Looze et al., “Exoskeletons for industrial application and their potential effects on physical work load,” Ergonomics, vol. 0139, pp. 1–11, 2016.

[3] S. Toxiri et al., “Rationale, Implementation and Evaluation of Assistive Strategies for an Active Back-Support Exoskeleton,” Front. Robot. AI, vol. 5, p. 53, 2018.

Abstracts Poster

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Influence of the Facet Joints on the Mechanical Behaviour of the Intervertebral Disc: the Numerical and Experimental Analysis

Żak M, Szkoda-Poliszuk K, Pezowicz C

Dept. of Biomedical Engineering, Mechatronics and Theory of Mechanisms, Wroclaw University of Science and Technology, Wroclaw, Poland

Many studies confirm the key role of the intervertebral disc (IVD) in the proper functioning of the spine. However, there is no analysis indicating the changes that occur inside the IVD depending on the support system. The presented studies are an innovative way of describing the influence of complex complex cyclic loads and lack of support on joint processes on the formation of pathological changes and their influence on the mechanical properties of the IVD.

Experimental studies were carried out on an animal model of the domestic pig lumbar functional spinal unit, divided into groups: control, intact and without posterior column (obtained by means of highly cyclical fatigue loads).The mechanical properties of the annulus fibrosus (AF) were determined on the basis of a uniaxial tension test.

The value of failure stress (σ_{UTS}) of the AF in the group without posterior column was lower than in the control group, with statistically significant differences were found only in the anterior part of the IVD (Fig. 1). The microscopic analysis showed that the pathological changes occurred mainly in the AF lamellas in the posterior part of the IVD. The main changes concern the penetration of nucleus pulposus between the inner lamellas of the AF and the heterogeneity in adjacent lamellas caused by delamination.

Finite element analysis were carried out on a spine model in order to assess the changes occurring in individual layers of the AF of IVD for different load systems (compression, flexion and hyperextension). The numerical simulations showed that in the physiological group all layers of the AF decrease their height regardless of the applied load, emphasizing the outside of the IVD (Fig. 1). Removal of the posterior column from the motion segment increases its mobility and disruption of the load transfer system, affecting the emphasizing of the AF lamellas inside the IVD.

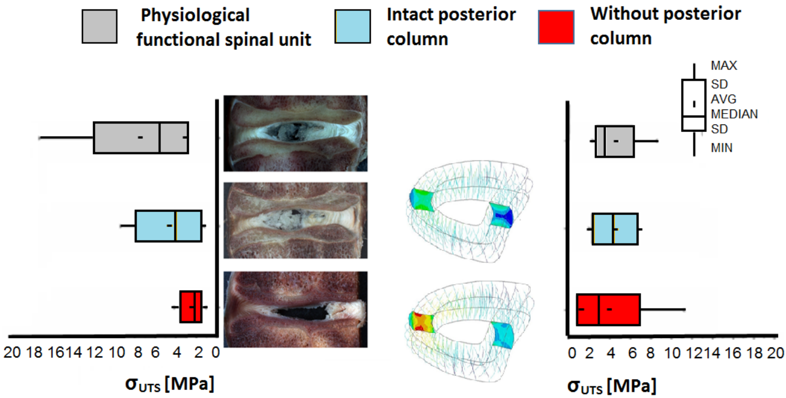


Fig. 1: Influence of the articular triad of mechanical and structural properties of the annulus fibrosus in experimental and numerical research.

Beyond Preload - The Replication of Six-axis In-Vivo Load Data using a Spine Simulator

Holsgrove TP[a,b]

[a]*College of Engineering, Mathematics and Physical Sciences, University of Exeter, Exeter, UK*

[b]*Centre for Orthopaedic Biomechanics, Department of Mechanical Engineering, University of Bath, Bath, UK*

The accurate replication of in-vivo loads is critical in order to understand natural spine biomechanics, and the efficacy of new medical devices. The stiffening effect of compressive loads on the spine is well documented [1]. However, there remains a lack of consistency in how preloads are applied, if they are applied at all, which can significantly affect spinal kinematics [2, 3]. The aim of this research was to assess the ability of a spine simulator to replicate in-vivo loads measured using an instrumented vertebral body replacement [4, 5].

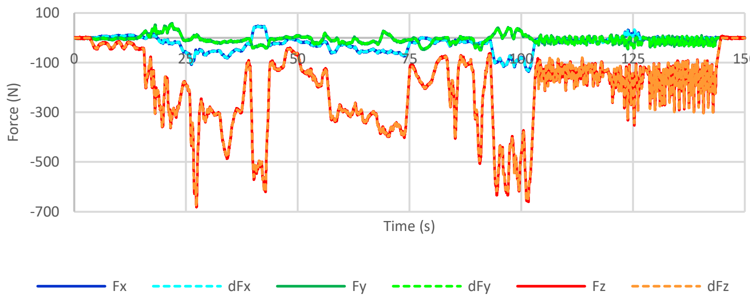
A six-axis spine simulator capable of six-axis load control was used for testing [6]. The Orthoload database [7] was used to obtain six-axis load data from the vertebral body replacement of a single patient during sixteen activities (Table 1). These data were compiled into a load demand file, with a 1 second transition between activities. Test were completed three times using a single standardised synthetic spring specimen [8]. The mean root mean squared (rms) error was calculated for forces and moments in each axes over the duration of each test.

The spine simulator accurately replicated force demand signals in all three axes (Fig. 2a), with mean±sd rms errors of 1.35±1.46 N, 1.53±1.69 N, and 6.50±7.52 N in the x, y, and z axes respectively. A low error of 0.10±0.08 Nm was also achieved in axial torque (Fig. 2b). However, whilst the load profile in flexion-extension and lateral bending reflected the demand signal, the errors of 0.42±0.54 Nm and 0.53±0.61 Nm for the x and y axes respectively were relatively large (Fig. 2b).It may be possible to minimise these errors through the introduction of real-time load transformation, and improved signal filtering. Nevertheless, this research provides a valuable step towards the replication of in-vivo loading in the in-vitro environment.

[1] Stokes et al, 2003. Journal of Electromyography and Kinesiology, 13(4): p397-402.
[2] Bennett et al, 2015. International Journal of Spine Surgery, 9: p33.
[3] Bell et al, 2015. Journal of Biomechanics, 49(2): p167-172.
[4] Rohlmann et al, 2014. Gait & Posture, 39(2): p750-755.
[5] Rohlmann et al, 2014. PLoS One, 9(5): pe98510.
[6] Holsgrove et al, 2017. Medical Engineering & Physics, 41: p74-80.
[7] Bergmann, G. (Ed.), Charité Universitaetsmedizin Berlin (2008) "Orthoload". Retrieved 02/01/2019 from https://orthoload.com.
[8] Holsgrove et al, 2018. Journal of Biomechanics, 70: p59-66.

Activity	Description	Data file
1	Supine, relaxed	WP1_200110_1_15.AKF
2	From lying to sitting	WP1_160408_1_93.AKF
3	Standing up no support	WP1_200110_1_167.AKF
4	Standing, relaxed	WP1_101210_1_16.AKF
5	Standing, upper body flexion	WP1_250907_1_30.AKF
6	Standing, upper body extension	WP1_050907_1_22.AKF
7	Standing, upper body right lateral bending	WP1_301107_1_5.AKF
8	Standing, upper body left lateral bending	WP1_101210_1_142.AKF
9	Standing, upper body left axial rotation	WP1_101210_1_144.AKF
10	Standing, upper body right axial rotation	WP1_050907_1_15.AKF
11	Putting 3 kg weight in a cupboard at head level	WP1_050907_1_103.AKF
12	Putting 3 kg weight in a cupboard at hip level	WP1_050907_1_87.AKF
13	Standing, sweeping floor	WP1_200110_1_84.AKF
14	Walking	WP1_140307_1_107.AKF
15	Walking upstairs	WP1_140307_1_123.AKF
16	Walking on a treadmill at 4 km/h	WP1_140307_1_132.AKF

Table 1: Orthoload load data [7] used in the present study. Activities were completed in the order shown.



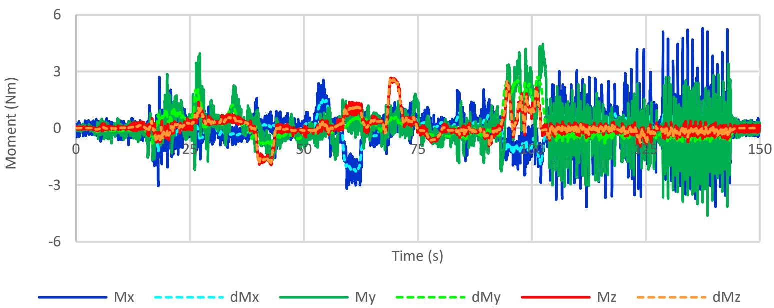


Fig. 1: Actual and demand (denoted d) data for forces (a) and moments (b) during a test of sixteen activities. Mx and My errors were substantially higher than other loads, particularly during the relatively high frequency loading rates of walking activities which occurred after approximately 100 seconds.

Two-level Fusion Versus Topping-off Technology based on Coflex in the Treatment of Lumber Degenerative Disease: a Biomechanical Effect Comparison

Sun XY, Kong C, Lu SB

Dept. of Orthopedics, Xuanwu Hospital, Capital Medical University, Beijing, China

The aim of this study is to compare the biomechanical effect of interspinous dynamic stabilization adjacent to single-level fusion with two-level fusion on range of motion (ROM) of the transition segment and the upper adjacent segment.

Eight fresh human cadaveric lumbosacral (L1–S1) spines were utilized in the following sequence: A) intact spine (ISP); B) single-level fixation in L5/S1 (SLF); C) SLF + dynamic adjacent segment fixation in L4/5 (DLF); D) two-level fixation in L4–S1 (TLF). ROM at L3/4 and L4/5 were recorded and calculated.

Under flexion/extension, the mean ROM of L3/4 increased in the order of SLF, DLF and TLF. Compared with ISP, the ROM of L3/4 after SLF and DLF showed a tendency to increase, but the difference was not significant ($P>0.05$). Compared with ISP, the ROM of L3/4 after TLF showed a significant increase ($P<0.001$). Under lateral bending and axial rotation, L3/4 ROM after TLF also showed a significant increase ($P<0.001$). L4/5 ROM after SLF significantly increased under flexion/extension ($P<0.05$), while both DLF and TLF significantly decreased L4/5 ROM ($P<0.05$). L4/5 ROM after SLF and DLF showed no significant change under lateral bending and axial rotation, while TLF significantly decreased L4/5 ROM under these conditions.

Fusion combined with Coflex can stabilize the transition segment, and restrict flexion and extension in that segment, while having a less significant effect on the ROM of adjacent segments compared with two-level fixation.

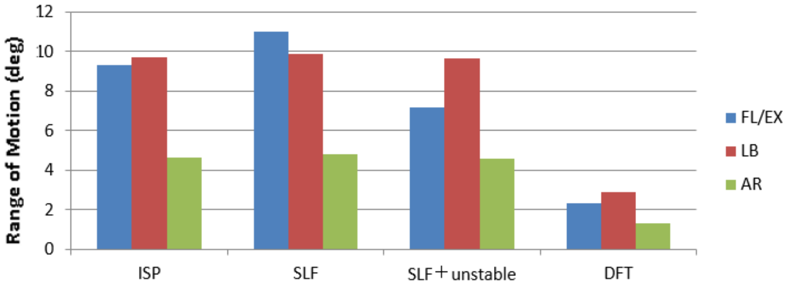


Fig. 1: L3/4 ROM of the four groups under flexion/extension, lateral bending and axial rotation; ISP, intact spine; SLF, single-level fusion in L5/S1; DFT, single-level fusion in L5/S1 and dynamic fixation in L4/5; TLF, two-level fixation in L4/5 and L5/S1.

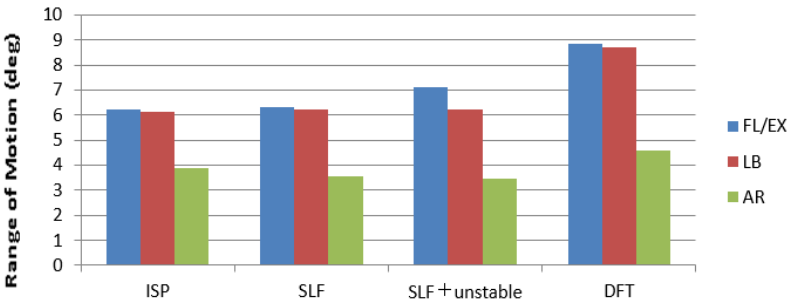


Fig. 2: L4/5 ROM of the four groups under flexion/extension, lateral bending and axial rotation; ISP, intact spine; SLF, single-level fusion in L5/S1; DFT, single-level fusion in L5/S1 and dynamic fixation in L4/5; TLF, two-level fixation in L4/5 and L5/S1.

Sublaminar Tape as Alternative and Addition to Pedicle Screws in Spinal Surgery

Doodkorte RJP[a], Roth AK[a], van Rietbergen B[a,b], Arts JJ[a,b], Lataster A[c], van Rhijn LW[a], Willems PC[a]

[a]*Dept. of Orthopaedic Surgery, Research School CAPHRI, Maastricht University Medical Center, Maastricht, The Netherlands*
[b]*Orthopaedic Biomechanics, Dept. of Biomedical Engineering, Eindhoven University of Technology, Eindhoven, The Netherlands*
[c]*Dept. of Anatomy and Embryology, Maastricht University Medical Centre, Maastricht, The Netherlands*

Complications after spinal fusion surgery are common, with implant loosening occurring in up to 50% of osteoporotic patients. Pedicle screw fixation strength reduces as a result of decreased trabecular bone density, whereas sublaminar wiring is less affected by these changes. Therefore, radiopaque sublaminar tape is expected to have a pullout strength similar to pedicle screws, and it might be feasible as screw reinforcement. Furthermore, tape could result in a gradual transition to reduce implant loosening. The objective of this study is to test this hypothesis in a novel experimental setup in which a cantilever bending moment is applied to individual human vertebrae.

Thirty-eight human cadaver vertebrae were stratified into four different groups: ultra-high molecular weight polyethylene sublaminar tape (ST), pedicle screw (PS), metal sublaminar wire (SW) and pedicle screw reinforced with sublaminar tape (PS+ST). The vertebrae were individually embedded in resin, and a cantilever bending moment was applied bilaterally through the spinal rods using a universal materials testing machine. This cantilever bending setup closely resembles the loading of fixators at transitional levels of spinal instrumentation (Fig. 1).

The pull-out strength of the ST (3563 ± 476N) was not significantly different compared to PS, SW or PS+ST. The PS+ST group had a significantly higher pull-out strength (4522 ± 826N) compared to PS (2678 ± 292N) as well as SW (2931 ± 250N) (Fig. 2).

The pull-out strength of the sublaminar tape was similar to pedicle screws, indicating that sublaminar tape could be an effective stand-alone spinal fixation method. The higher failure strength of PS+ST compared to PS demonstrates the feasibility of pedicle screw augmentation with sublaminar tape to reduce the incidence of screw pull-out. Further testing with multi-level spinal segments is necessary to demonstrate a beneficial ‘gradual transition’ effect of using sublaminar tape at the construct ends.

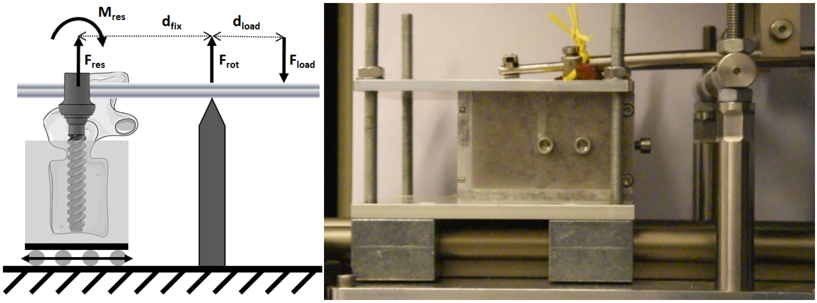


Fig.1: Experimental setup designed to mimic *in vivo* loading of spinal fixation members. By applying a compressive force onto an instrumentation rod, positioned over a hinge point, a pull-out force and moment are applied to the fixation member. An excess of shear forces was avoided by mounting the specimen on a sliding surface.

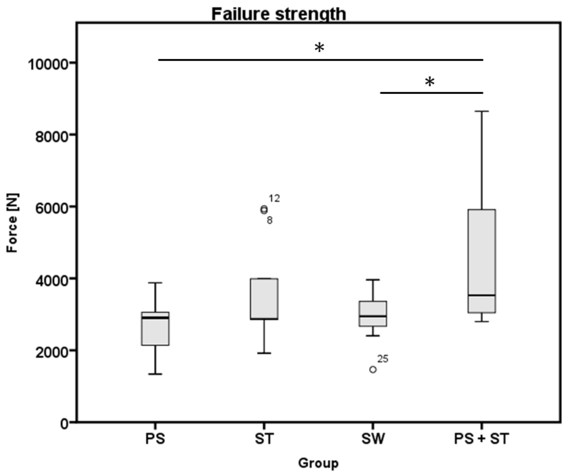


Fig. 2: The failure strength of pedicle screw constructs reinforced with sublaminar tape (PS + ST) is significantly higher compared to stand alone pedicle screws (PS) or metal sublaminar wires (SW). The pullout strength of sublaminar tape (ST) was comparable to PS and SW.

Effects of the Nucleus Migration during Forward Flexion on the Biomechanics of the L4-5 Functional Spinal Unit

Liu T[a], Khalaf K[b], El-Rich M[a]

[a]*Dept. of Mechanical Engineering, Khalifa University, Abu Dhabi, UAE*
[b]*Dept. of Biomedical Engineering, Khalifa University, Abu Dhabi, UAE*

Migration of the nucleus pulposus (NP) with changing posture within the intervertebral disc (IVD) continues to be debated. While some studies report apparent movement of the NP during flexion and extension of the spine [1], others indicate that the NP migration with changing posture (from supine to sitting and standing) in fact reflects deformation of the length of the NP, which in turn depends on the posture and the magnitude of the load [2].

This study investigated the impact of NP migration during flexion on spinal loading.

A previously validated tool, which combines musculoskeletal modeling of the trunk with Finite Element (FE) modeling of the lumbosacral spine [3], was used to simulate 60°-flexion. The muscle forces predicted by the musculoskeletal model were applied, along with gravitational forces, to the L4-5 segment of the FE model. Three locations within the NP, namely: the center of the IVD (Model-0); 1.5mm posteriorly (Model-1.5P); and 2.7mm posteriorly (Model-2.7P) from the disc center, were studied (Fig. 1). The locations were calculated based on the anterior-posterior disc distance (29mm) and the *in-vivo* migration percentage (6%~9%) with respect to anterior-posterior disc length during 60°-flexion angle [1]. The range of motion of the modified L4-5 segments fell within the *in-vivo* range. Positions of the joints simulating the discs in the musculoskeletal model were kept unchanged.

Posterior migration of the NP reduced the intradiscal pressure by up to 28%. The magnitude predicted by Model-2.7P approached the *in-vivo* value. Compressive forces in the disc also decreased from 1370N in Model-0 to 1270N in Model-1.5P, and then to 1210N in Model-2.7P. Our study confirms that spinal FE models should account for the NP migration when studying forward flexion. Ongoing work includes the effect of NP migration on the entire lumbar spine.

[1] Fennell, A.J., Jones, A.P., Hukins, D.W., 1996. Migration of the nucleus pulposus within the intervertebral disc during flexion and extension of the spine. Spine 21, 2753–2757.

[2] Nazari, J., Pope, M.H., Graveling, R.A. Reality about migration of the nucleus pulposus within the intervertebral disc with changing postures. Clinical Biomechanics 27, 213-217.

[3] Liu, T., Khalaf, K., Naserkhaki, S., El-Rich, M., 2018. Load-sharing in the lumbosacral spine in neutral standing & flexed postures - A combined finite element and inverse static study. Journal of Biomechanics 70, 43-50.

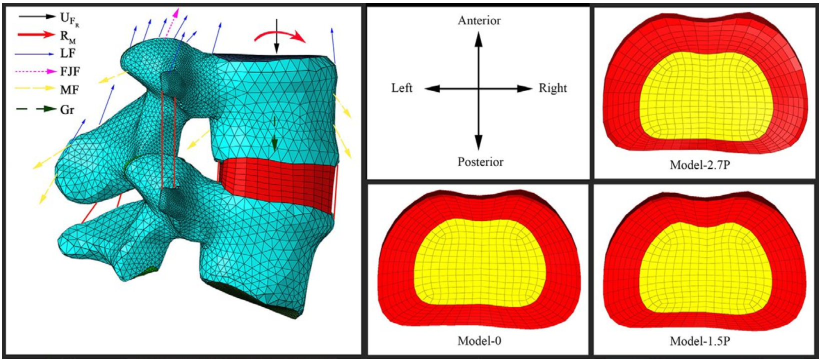


Fig. 1: FE models with different NP migration.

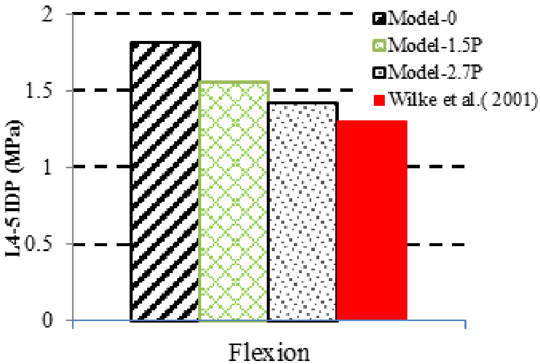


Fig. 2: Effects of NP migration during forward flexion on IDP.

Sensitivity of Musculoskeletal Model Predictions in Neutral Standing and Forward Flexion Postures to Center of Rotation Location

Liu T[a], Khalaf K[b], El-Rich M[a]

[a]*Dept. of Mechanical Engineering, Khalifa University, Abu Dhabi, UAE*
[b]*Dept. of Biomedical Engineering, Khalifa University, Abu Dhabi, UAE*

Musculoskeletal (MSK) modeling is a powerful tool that uses kinematics of the human body to predict muscle forces and joint loads. Most published MSK models of the trunk simplify the intervertebral discs as spherical joints located at the mean center of rotation (CoR) based on *in-vivo* measurements [1]. The consequences of this assumption on the fidelity of prediction of muscle forces and joint loads remain unclear.

This study investigated the effects of the CoR location on muscle forces, joint forces and lumbar spine kinematics in neutral standing (NS) and 60° flexed postures (FLX).

A previously validated MSK model of the trunk, including the skull, upper arms, thorax, and lumbosacral spine (average height of 168cm and weight of 70kg), was employed. The CoR locations were generated using the Latin hypercube sampling method based on *in-vivo* measurements [1]. A total of 482 simulations were completed, and the predicted spinal and muscle forces were calculated and analyzed.

The results demonstrate that the mean CoR does not predict mean muscle and spinal forces (Fig. 2), and that the CoR location significantly influenced muscle forces, especially in FLX where the variation in the Internal Oblique force was as high as 14 times the minimum value (13.9N). In NS, the maximum force in the Psoas Major was 9 times the minimum force. Compressive force at the L4-5 level varied between 550N and 1050N in NS, and between 800N and 1900N in FLX. The direction of the shear force changed at L4-5 in NS (-50N~150N), where the magnitude varied between 52N and 360N in FLX.

Accurate prediction of muscle forces and spinal loads during FLX using MSK modeling requires personalized CoRs, in addition to the anthropometric data. Ongoing work includes investigating sensitivity of the entire lumbar spine kinematics to the CoR.

[1] Pearcy, M.J., Bogduk, N., 1988. Instantaneous axes of rotation of the lumbar intervertebral joints. Spine 13, 1033–1041.

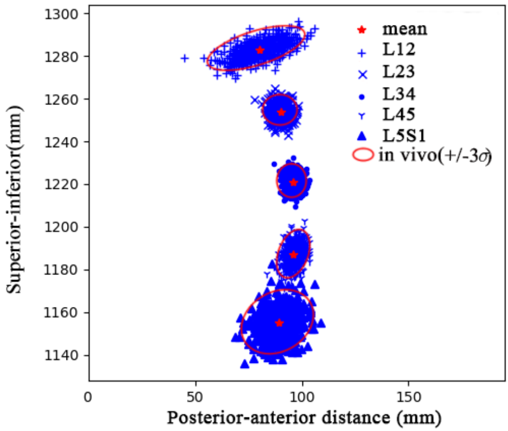
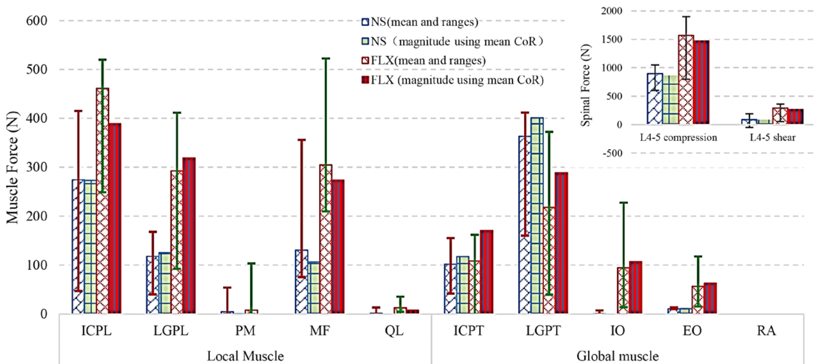


Fig. 1: Sampled locations of the CoR in the MSK model compared to the *in-vivo* range.



Global Muscles: RA-Rectus Abdominis, IO-Internal Oblique, EO-External Oblique, ICPT-Iliocostalis Lumborum Pars Thoracic, LGPT-Longisimus Thoracis Pars Thoracic

Local Muscles: ICPL-Iliocostalis Lumborum Pars Lumborum, LGPL-Longisimus Thoracis Pars Lumborum, PM-Psoas Major, MF-Multifidus, QL-Quadratus Lumborum

Fig. 2: Mean values and ranges of the predicted muscle forces.

A new Method for Validation of an Individual Forward Dynamics Model of the Lumbar Spine

Damm N[a], Aiyangar A[b], Gruber K[a]

[a]*MTI Mittelrhein, Institute for Medical Engineering and Information Processing, University of Koblenz-Landau, Campus Koblenz, Koblenz, Germany*

[b]*Mechanical Systems Engineering, EMPA (Swiss Federal Laboratories for Material Science and Technology), Duebendorf, Switzerland*

Medical diagnostics of damage and degeneration of the lumbar spine has mainly been based on imaging techniques such as CT, MRI and X-rays, which are usually recorded only static and in lying position, particularly. The possibility of a dynamic representation of the entire 3D movement sequence of the individual patient lumbar spine would be a decisive advance for improved diagnostics. This can be realized by an individual 3D dynamics-based computational model including all essential passive structures and muscles. We propose a novel method for validation of such a model using dynamic stereo X-ray (DSX) imaging.

Based on a 3D CT scan and an X-ray film of a lumbar spine extension movement, a personalized Multi Body Simulation (MBS) computer model was constructed by segmenting the individual vertebral surfaces and transferring them to the modelling software. The model includes the essential passive structures (IVD, ligaments and facet joints) as well as the modelled musculature of m. psoas major and mm. multifidi based on Hill's muscle equations, which allow for forward dynamics through appropriate stimulation. Matching the 3D vertebral surfaces to the 2D X-ray images provided the target positions for the vertebrae of the computer model for each individual image of the X-ray film. Thus, the model was validated by superposition of simulated motion and X-ray film with careful adaptation of parameters of passive structures and appropriate stimulation of the musculature.

When starting from the upright position, the simulation showed an optimal match of the movement of the computer model with the movement in the X-ray film. The validated computer model was used to calculate the loads in the passive structures and in the muscle force elements of the lumbar spine.

The method opens up possibilities for determining biological parameters of the living individual, which offers advantages compared to cadaver experiments.

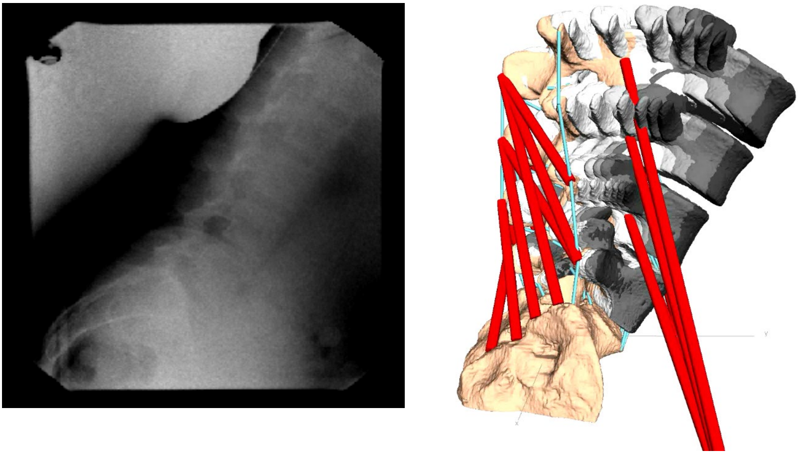


Fig. 1: (a) Motion of the lumbar spine recorded by a dynamic stereo X-ray imaging system.
(b) Simulation of the individual forward dynamics mbs model with muscle forces.

Single Rigid Segment versus Multi-segmental Approach for the Analysis of the Lumbar Spine in Low Back Pain

Papi E[a,b], Bull AJ[b], McGregor AH[a]

[a]*Dept. of Surgery and Cancer, Imperial College London, London, UK*
[b]*Dept. of Bioengineering, Imperial College London, London, UK*

Spine kinematics analysis provides useful information for the understanding of low back pain (LBP). Whilst the approach, over the years, has been to represent the lumbar spine as one rigid segment, recent studies have highlighted the importance of a multi-segmental approach. This study investigated if differences between people with and without LBP can be found when modelling the lumbar spine as a single segment versus a multi-segmental approach.

Twenty participants with and twenty without non-specific LBP were recruited. Participants performed 3 trials of walking, sit-to-stand, and lifting a 5kg box. A 3D motion capture system was used for data collection. Markers on L1, L3 and L5 were used to define upper and lower lumbar spine segments and the rigid lumbar segment. 3D angular kinematics were calculated for each segment. Independent t-tests were used to compare the lumbar segments' range of motions (ROMs) between groups. Moreover, statistical parametric mapping (SPM) was used to assess group differences between mean joint angles over each entire task cycle. Significance was set at 0.05.

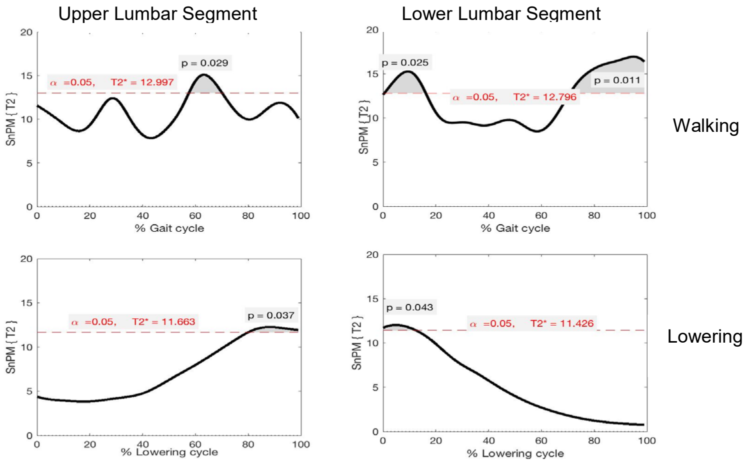
The upper lumbar segment ROMs were significantly different between groups during walking (flexion), sit-to-stand (lateral flexion) and lifting (flexion) ($p<0.05$; Table 1). Lower lumbar flexion and rotation ROMs were significantly different during walking and lifting respectively ($p<0.02$; Table 1). The lumbar segment ROM showed significant differences only during lifting ($p<0.01$; Table 1). SPM analysis revealed no differences for the rigid lumbar segment in all tasks whereas regional differences were observed in the 3D angular components of the upper and lower lumbar segments during walking and lifting at different intervals over the task cycles (Fig. 1).

The findings demonstrate that analysing the lumbar segment as one rigid segment leads to important information on spine kinematics being lost. Furthermore, analysing only ROM values leads to differences in movement strategy being overlooked or overestimated.

Table 1: Mean ROMs (° ± standard deviation) in the 3 anatomical planes for Healthy and LBP groups for all segment analysed during the tasks performed. Lifting task was divided into lowering and picking phase. Significant values are in bold font.

		Lumbar Segment			Lower Lumbar Segment			Upper Lumbar Segment		
		F	T	S	F	T	S	F	T	S
Walk	Healthy	9.18 (±4.1)	5.22 (±2.9)	5.44 (±3.1)	6.76 (±2.2)	5.13 (±2.6)	**5.59 (±3.1)**	7.36 (±7.5)	6.74 (±4.5)	**6.24 (±4.7)**
	LBP	9.83 (±4.7)	5.27 (±2.6)	6.72 (±3.4)	7.49 (±2.7)	4.96 (±2.1)	**7.79 (±3.8)**	8.05 (±6.1)	9.46 (±9.7)	**9.61 (±8.5)**
Sit-to-stand	Healthy	2.71 (±1.0)	2.75 (±1.4)	10.56 (±5.7)	2.71 (±1.0)	2.73 (±1.4)	8.85 (±4.8)	**1.97 (±1.0)**	2.16 (±1.2)	11.72 (±5.2)
	LBP	2.82 (±1.6)	2.77 (±1.0)	10.78 (±7.6)	2.55 (±1.3)	2.77 (±1.0)	9.56 (±7.0)	**3.65 (±2.3)**	2.51 (±1.7)	11.38 (±5.9)
Lowering	Healthy	1.89 (±1.1)	**1.98 (±1.1)**	**15.79 (±7.2)**	1.78 (±0.9)	2.12 (±1.1)	10.58 (±7.1)	2.28 (±1.3)	1.93 (±1.1)	**18.15 (±4.9)**
	LBP	2.50 (±1.4)	**2.72 (±1.2)**	**11.63 (±5.7)**	2.60 (±1.5)	2.71 (±1.3)	9.42 (±5.4)	3.19 (±2.2)	2.07 (±0.9)	**13.49 (±6.7)**
Picking	Healthy	1.95 (±1.1)	2.05 (±0.8)	16.05 (±7.9)	**1.76 (±0.9)**	2.10 (±1.0)	11.13 (±7.3)	2.27 (±1.0)	1.91 (±1.1)	16.41 (±6.7)
	LBP	2.52 (±1.3)	2.77 (±1.3)	14.04 (±7.2)	**2.80 (±1.5)**	2.70 (±1.1)	11.46 (±7.1)	3.15 (±1.9)	2.02 (±1.2)	14.55 (±7.7)

F: Frontal plane; T: Transverse plane; S: Sagittal plane



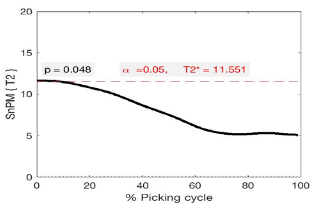


Fig. 1: SPM Hotellings T^2 test analysis results for the 3D angular components of the lower and upper lumbar segments. The horizontal dotted line indicates the critical threshold and the grey areas depict where the groups are statistically significant different over the task cycle.

The Workload on One's Low Back during Dishwashing

Zhang H and Zhu W

Shanghai Institute of Applied Mathematics and Mechanics, Shanghai University, Shanghai, China

Dishwashing increases the workload on one's low back and it has been recognized as a risk factor of low back pain. However, the details of the load on lumbar spine and its surrounding tissues in the washing process is little known yet. This study aimed to calculate the loads on the lumbosacral spine containing all attached tissues during dishwashing and to investigate the effects of the body height and the height of sink on spinal load. A 3D nonlinear finite element model of lumbosacral vertebrae (L1-S1) with calibrated material properties was established and validated. All tissues attached to it, including the spinal cord muscles, ligaments, fat, etc., have also been developed and homogenization simplified. In washing, people have to keep the upper-body leaning forward for long periods. The flexion angles mainly depended on the height of the sink and the body height. Therefore, the loads acting on the lumbar spine can be obtained from the measurement and calculation data of the upper-body bending. The intradiscal pressure and the stress on several tissues were calculated for upper-body flexion angles between 0 deg and 30 deg in steps of 10 deg during dishwashing. The intradiscal pressure and the stress of several tissues were strongly affected by the upper-body flexion angles. An evident nonlinear relationship between upper-body flexion angle and internal load of the low back exists. On average, the load acting on the spine and local attached tissues for bending forward 30 deg during washing dishes was 300% of the value for 0 deg. Low back pain was mainly caused by the high load of the upper-body flexion position during dishwashing. Therefore, we suggested that the absolute difference between the height of kitchen sink and the body height should be less than 80 cm to avoid excessive forward bending posture in dishwashing process.

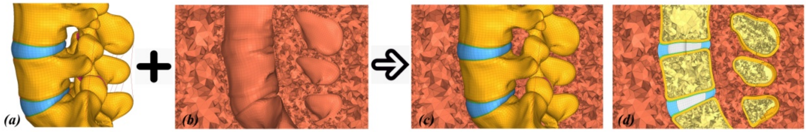


Fig.1: Finite element model for human low back. (a) Finite element model for lumbosacral vertebrae L4-S1. (b) Finite element model for several tissues around the lumbosacral spine. Only the right half of the tissues were showed. (c) Finite element model for the lumbosacral spine wrapped by surrounding tissues. (d) Finite element model for low back in right half part.

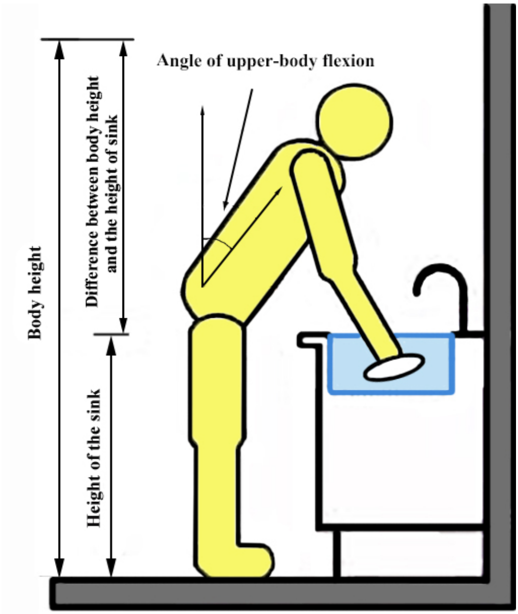


Fig. 2: The angle of upper-body flexion during dishwashing.

Smart Rotational Spine Protector (RSP) for Sport and Rehabilitation

Rafolt D[a,b], Saier T[a,c], Meixner L[a], Stiegler R[a], Russ D[a]

[a]*Edera-Safety GmbH&CoKG, Lebring, Austria*
[b]*Center for Medical Physics and Biomed. Engineering, Med. Univ. Vienna, Austria*
[c]*University of Applied Science, Carinthia, Austria*

State-of-the-art back protectors used in sports only protect against direct impact traumas.

The majority of heavy spine injuries – paraplegia at worst – are based on extensive movements exceeding the physiological normal range – especially over-rotation of the spine.

We developed a new generation of spine protectors that acts as an extension (B2B model) for state of the arte back protectors. A system of diagonal belts connects hip and shoulder regions allowing normal movements without restrictions but stops rotation if a certain limit is reached. Due to a special arrangement of the belts rotation limits are unaffected form flexion and extension of the upper body.

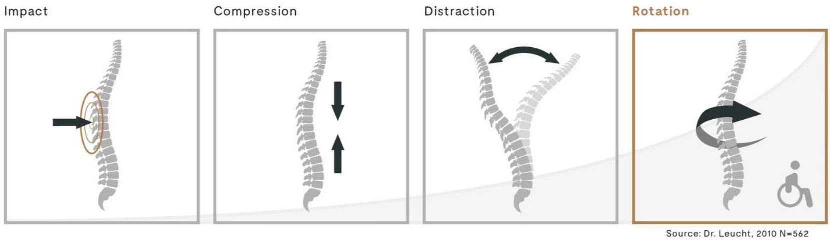
Prototypes were equipped with IMUs (inertia measurement units) containing accelerometers and gyrometers to record 3-dimensional movements of hip and shoulder separately. Simple subtraction leads to relative date that support us with the process of rotation and 3D-flexion of the spine. To record the blocking load during an accident force sensors are integrated in the diagonal straps.

For evaluation of the efficiency of both – state-of-the-art protectors and the new RSP device respectively we developed a rotational test stand driven by a high dynamic pneumatic muscle that is able to record rotation angel and input torque to generate compliance curves of a torso dummy equipped with various protectors.

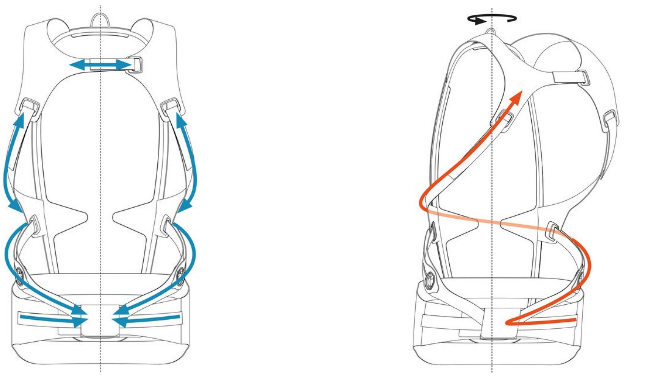
A modification of the test stand allows to carry cadaver spines and 3D-printed spines respectively in order to get data about the ultimate strength of spines on the one hand and to get knowledge for 3D-printing materials for dummies on the other hand.

Field measurements with mountain bike test drives and inhouse test stand measurements have proven, that there is no restriction during normal use in sports but an effective restriction of rotation in extreme situations.

Using the new smart RSP system without the protective back plate of classical back protectors but with adjustable limits allows the application in the field of rehabilitation e.g. after injuries and surgeries. Step by step extension of the strap limits allows efficient exercises in the rehabilitation process and protect the patient at the same time for overloads.



Only 3,9% of impact traumas are responsible for spinal cord injuries.



Relaxed state: Slight tension in the system during normal usage means a “snug” fit around the body.

Twisted State: During an accident the RSP System tightens around the body impeding excessive rotation.

dietmar.rafolt@medunivien.ac.at
www.meduniwien.ac.at
d.rafolt@edera-safety.com
www.adamsfour.at

Biomechanical Performance of Rigid and Semi-Rigid Fixation Subject to Static and Cyclic Loading

Khalaf K[a], Nikkhoo M[b], Wang JL[c], El-Rich M[d], Hebela N[e], Almansouri S[e,] Parnianpour M[f]

[a]*Dept. of Biomedical Engineering, Khalifa University, Abu Dhabi, UAE*
[b]*Dept. of Biomedical Engineering, Science and Research Branch, Islamic Azad University, Tehran, Iran*
[c]*Institute of Biomedical Engineering, National Taiwan University, Taipei, Taiwan*
[d]*Dept. of Mechanical Engineering, Khalifa University, Abu Dhabi, UAE*
[e]*Orthopaedic Spine Surgery, Neurological Institute, Cleveland Clinic Abu Dhabi, Abu Dhabi, UAE*
[f]*School of Mechanical Engineering, Sharif University of Technology, Tehran, Iran*

While posterior lumbar fusion (PLF) using titanium (Ti) rods remains the standard treatment modality for many lumbar degenerative conditions, its adverse effects, including accelerated degeneration of adjacent segments, are well known [1]. Several dynamic and semi-rigid immobilization systems have therefore been introduced towards enhanced clinical results [2]. This study aims to analyze the biomechanical performance of Ti and Polyetheretherketone (PEEK) rod constructs during static and cyclic loading.

Using a poroelastic osteoligamentous FE model of the human spine [3], fusion was simulated at L3-L4 level, and the biomechanics of adjacent levels were studied. Two instrumented models with Ti and PEEK rods (Fig. 1) were developed and compared after 8h rest (200N), following 16h cyclic compressive loading of 500-1000N (40 and 20min, respectively). In addition, different movements (i.e., flexion, extension, lateral bending and axial rotation) were simulated using 10Nm moment before and after cyclic loading (Fig. 2).

The intact FE model results were well comparable with in-vitro [4], as well as, FE studies [5]. The PEEK construct demonstrated slightly increased range of motion (ROM) at the instrumented level, but decreased ROM at adjacent levels, as compared to the Ti. During cyclic loading, disc height loss, fluid loss, axial stress, and collagen fiber strain in the adjacent IVDs were higher for the Ti construct when compared to intact and PEEK models. The Ti construct also exhibited higher average von-Mises stress in the screw-bone region as compared to the PEEK.

The main contribution of this work lies in providing a validated poroelastic FE model to analyze the fluid-solid interaction of IVDs in the lumbar spine for rigid and semi-rigid fixations. The results confirm differences in the poroelastic characteristics of adjacent discs for rigid (Ti) and semi-rigid (PEEK) constructs, and reveal the advantage of PEEK for decreasing the risk of adjacent level degeneration.

[1] Yoshihara et al., Spine J., 2013 (13); [2] Kurtz et al., European Spine J., 2013 (22); [3] Khoz et al., Iranian J. of Orthopaedic Surgery, 2018 (15); [4] Panjabi et al., J. Bone Joint Surgery, 1994(76); [5] Schmidt et al., J. Biomechanics, 2010 (43).

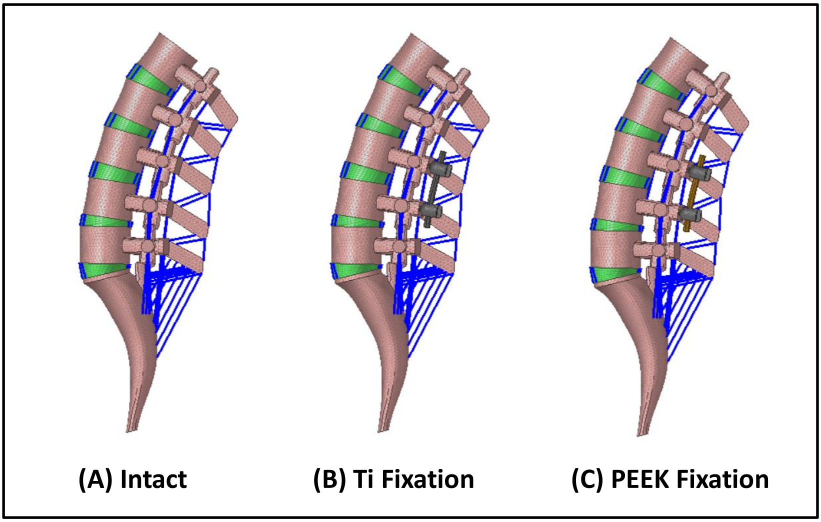


Fig. 1: Poroelastic finite element models for (A) intact, (B) Ti Fixation, (C) PEEK Fixation.

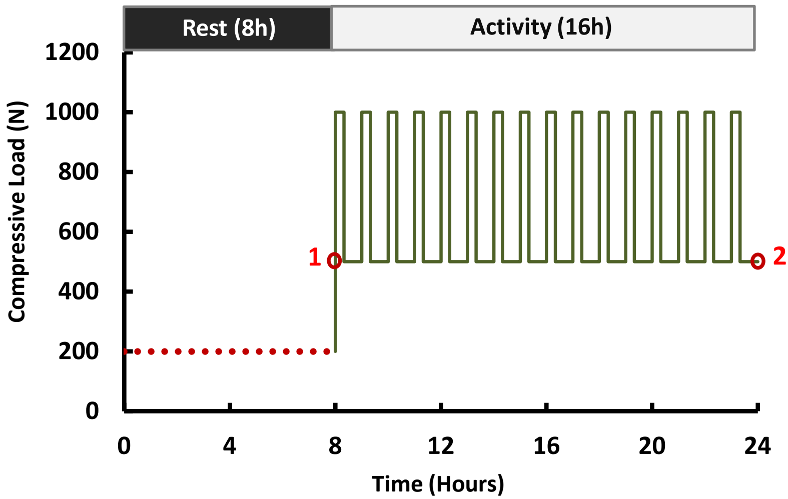


Fig. 2: Loading scenario of the compressive force (Flexion, extension, lateral bending and axial rotation moments of 10 N.m were applied at points 1 and 2).

Using SHARIF-HMIS Inertial Sensor for Measurement and Comparison of Kinematic Parameters in 3 Subgroups of STarT Back Screening Tool in Patients with Nonspecific Low Back Pain

Abedi M[b], Davoudi M[a,b], Shokouhyan SM[a,b], Manshadi FD[b], Narimani R[a], Hoviattalab M[a], Parnianpour M[a]

[a]*Laboratory of Wearable Technologies and Neuromusculoskeletal Research, Dept. of Mechanical Engineering, Sharif University of Technology, Tehran, Iran*
[b]*Physiotherapy Research Centre, School of Rehabilitation, Shahid Beheshti University of Medical Sciences, Tehran, Iran*

Validated Persian version of STarT divides the patients with chronic non-specific chronic low back pain into three subgroups of low, moderate, and high risk. The questionnaire covers social psychosocial aspects and has clinical application due to its easy use. Purpose of this study is to use the SHARIF-HMIS portable and repeatable new inertial sensor to measure motion in order to evaluate the kinematic indices in the three sub-groups defined by this questionnaire.

One hundred male patients by STarT were divided into three subgroups and were uniformly distributed in terms of age ($p=0.346$) and BMI ($p=0.829$) and without significant difference in pain measured VAS scale ($p=0.46$). The sensor was placed on the chest by straps and patients became familiar with test process. They reached the maximum flexion and returned to maximum extension at the maximum speed in sagittal plane for 15 seconds. Same process was performed in 15- and 30-degree rotation states to the left and right controlled by lines on the ground and an audio feedback control. Kinematic variables including maximum and average angular velocity, linear acceleration and jerk which is derived from angular acceleration, were extracted from sensor data and analyzed using repeated measure ANOVA.

The effect of grouping on maximum and average speed indices ($p= 0.196$, $p=0.253$), acceleration ($p = 0.69$, $p=0.061$) and Jerk ($p = 0.394$, $p=0.251$) were not significant. But with rotation on the transverse plane, a significant difference was made so that maximum and average speed and acceleration decreased and jerk increased with increasing asymmetry ($p <0.001$). Speed and acceleration in flexion were more than extension, that was inverse in jerk ($p <0.001$).

Grouping by STarT had no effect on kinematic indices in sagittal plane but showed significant effects in pre-rotated transverse angles of motion. STarT is simple and practical test but its kinematic effects depends on plane of motion. Further natural clustering and its relation to three level risk specified by Start should be explore.

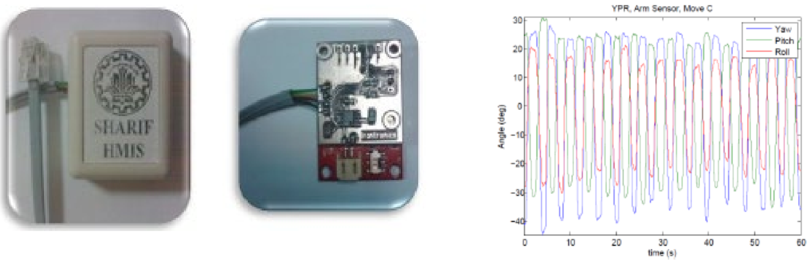


Fig. 1: SHARIF-HMIS inverter with Electronic board and an output sample.

Fig. 2: A person tested with Inertial sensor on the chest.

Risk for Fatigue-related Degeneration of the L5-S1 Disc among Persons with vs. without Unilateral Lower Limb Amputation

Hendershot BD[abc] and Bazrgari B[d]

[a]*DoD-VA Extremity Trauma and Amputation Center of Excellence, Bethesda, MD, USA*
[b]*Walter Reed National Military Medical Center, Bethesda, MD, USA*
[c]*Uniformed Services University of the Health Sciences, Bethesda, MD, USA*
[d]*University of Kentucky, Lexington, KY, USA*

Persons with lower limb amputation (LLA) commonly report low back pain (LBP) and perceive altered motions of the trunk/spine during activities of daily living as the primary contributor to its onset and/or recurrence [1]. Here, we expand on prior work linking such motions with elevated spinal loads [2], by estimating risk for long-term fatigue degeneration of spinal tissues (e.g., L5-S1 intervertebral disc). We hypothesized that repeated exposure to larger spinal loads among persons with vs. without LLA, as we reported in our earlier studies, would accelerate the risk for fatigue failure of the L5-S1 disc.

A non-linear, multiaxial fatigue damage model was used to estimate progression of fatigue damage at the L5-S1 disc over time [3, 4], with Von-Mises stresses calculated from prior estimates of compression and shear forces at the L5-S1 level during both (self-selected) walking and sit-to stand/stand-to-sit activities among persons with and without LLA [2, 5]. The fatigue damage estimations accounted for the combined effects of both activities and considering 5,000 (10,000) steps and 100 (120) sit-to stand/stand-to-sit repetitions per day for persons with (without) LLA.

The mean Von-Mises stresses experienced at the L5-S1 spinal level during each walking and sit-to-stand/stand-to-sit cycle were generally larger among persons with vs. without LLA (Fig. 1), resulting in an estimate of complete damage (i.e., $D_{mi}^{r} = 1$) among persons with LLA after ~8 years, in stark contrast to the estimated damage ($D_{mi}^{r} = 0.4$) among uninjured controls after 20 years (Fig. 2).

Repeated exposures to larger than normal spinal loads during two common activities of daily living accelerated damage evolution for the L5-S1 disc among persons with vs. without LLA, and thus additional work is needed to provide a predictive platform for evaluating specific clinical interventions that are ultimately intended to reduce the long-term burden and impact of LBP secondary to LLA.

[1] Devan et al, 2015 *Disability and Rehabilitation* 37: 873-883;
[2] Hendershot et al, 2018 *Journal of Biomechanics* 7-: 249-254;
[3] Chaboche and Lense 1988 *Fatigue fracture engineering material structure* 11:1-17;
[4] Motiwale et al 2018 *Advances in Mechanical Engineering* 10 (6): 1-16.
[5] Shojaei et al. 2018 *Clinical Biomechanics* (under review).

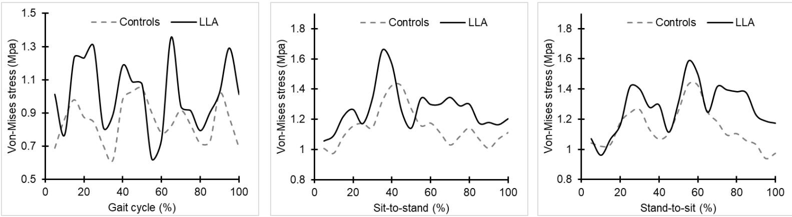


Fig. 1: Mean Von-Mises stresses at the L5-S1 disc during walking (left), sit-to-stand (middle), and stand-to-sit (right) activities, each depicted as a function of percent movement cycle, for persons with lower limb amputation (LLA) and uninjured controls. The required spinal loads for calculation of Von-Mises stresses were extracted from our earlier estimation of spinal loads during walking [2] and sit-to-stand and vice-versa [5] that were respectively involved 26 (26) and 10 (10) persons with (without) LLA.

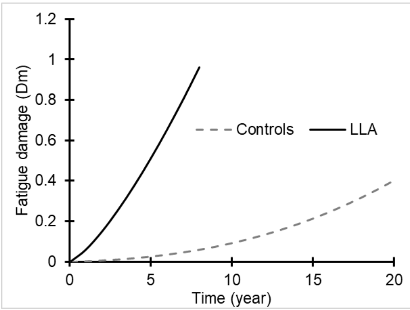


Fig. 2: Estimated damage evolution at the L5-S1 disc for persons with lower limb amputation (LLA) and uninjured controls. $D_{mi}^{r} = 1$ indicates complete fatigue damage.

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From the freeway A 100 take the exit Seestraße. Ample parking is available in the public parking garage at Seestraße 4. The garage is always open and costs 1 € for every full/partial hour or maximum 10 € per calendar day. The first 29 minutes are free. Guests who stay at the Virchow-Gästehaus have free parking included here.
On the campus the first 59 minutes are free and every hour afterwards costs 2 €. Disabled parking is available on the campus on Mittelallee.

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Registration for the workshop is required. Please contact Friedmar Graichen at: friedmar.graichen@charite.de

Registration fee for participants without oral presentation is required
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An invoice and confirmation of payment will be sent via electronic mail.

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The workshop language is English.

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Please prepare your presentation in MS Office PowerPoint up to 16:10 aspect ratio. A presentation notebook with Acrobat PDF Reader and PowerPoint 2016 will be provided. The use of personal notebooks will not be accepted, it may interrupt the flow of the program in the lecture hall. A laser pointer will be available at the speaker's podium in the lecture hall. A technical supervisor will help you.

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Please hand in your presentation on USB flash drive to our technical staff available in the room where the talk is scheduled, no later than 90 minutes before the beginning of the session. You may view and/or edit your presentation before.

Poster presentation
Panels of 123 cm height and 198 cm width will be available at the foyer for posters. Optimal poster size would be 120 cm height by 85 cm width (A0 portrait) for mounting two posters side by side on one panel. A poster should be self-contained and self-explanatory. Presentations should be kept simple and clearly visible from about 2 meters away with a balanced mix of text and graphics. Posters have to be installed at Thursday, July 4th.
We scheduled the poster session to Saturday, July 6th, 9:35-10:30 am.
Presenters are prepared for discussions in front of their posters.

Virchow Guesthouse Charité
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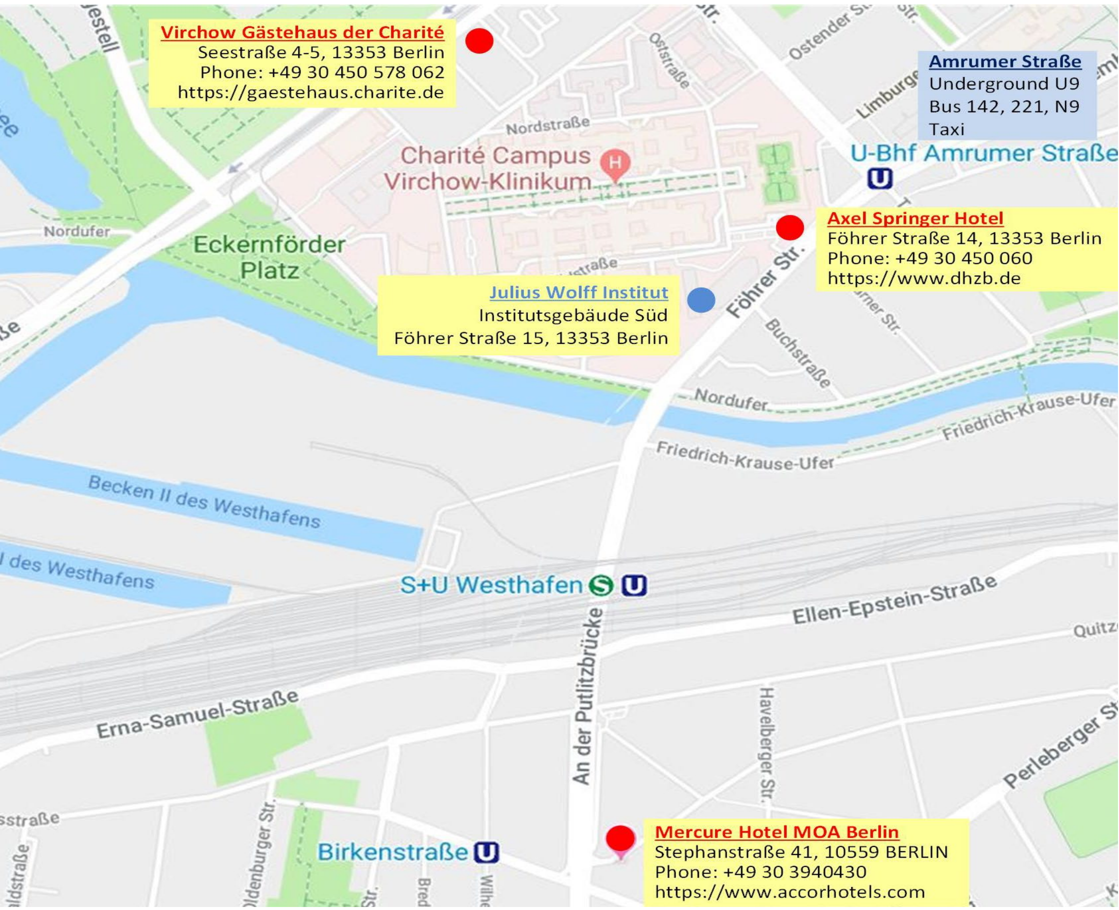
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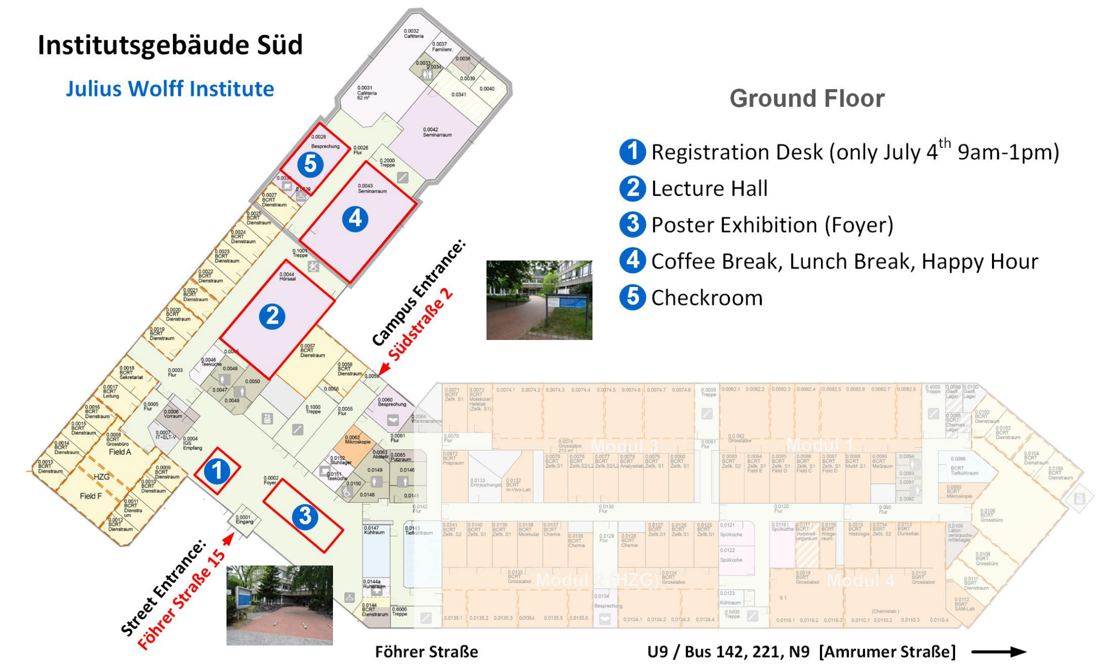
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