For the public defense of the PhD thesis of Arno Bisschop entitled:

Spinal stability following single level lumbar laminectomy: A biomechanical study of the human spine


The reception will be held afterwards in The Basket at the Vrije Universiteit campus.

Questions related to the PhD defense can be addressed to my paranymphs, Kees-Pieter Paul and Puck Vergroesen (arno.promotie@gmail.com).

I hope to welcome you at the ceremony and to meet you at the reception!

Best regards,

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Spinal stability following single level lumbar laminectomy: A biomechanical study of the human spine

About the author

Arno Bisschop was born in Hoorn, The Netherlands on February 12, 1987. He completed his secondary education at the Han Fortmann College (Heerhugowaard) in 2005. During his high-school period, he pursued a sportive career in speed skating and competed in multiple Dutch championships. In September 2005, he started with his bachelor in biomedical sciences at the VU university. After obtaining his propaedeutic examination, he commenced his bachelor in medicine in 2006. Due to his sportive career, he developed a keen interest for orthopedic surgery. As a result, he started his research at the department of orthopedic surgery at the VU university medical center within the Honours program. Later, this research evolved into a PhD position under supervision of prof. dr. B.J. van Royen, prof. dr. J.H. van Dieën and dr. I. Kingma. In the mean time, he completed his minor in entrepreneurship and started working as an anatomy teacher at the department of anatomy at the VU university medical center and as surgical assistant at the department of orthopedic surgery at Het Kennemer Gasthuis in Haarlem.

Besides his orthopedic and sportive ambitions, his interests in healthcare economics and healthcare based entrepreneurship also grew during this period. As a result, he co-founded the Medical Business Foundation and several start-up companies. Just recently he obtained his master in medicine with honours and his master in management, policy-analysis and entrepreneurship in health and life sciences. During the latter, he performed research on The Spine Clinic, a conceptual model of a privately held clinic focusing on value-based spine care, innovative techniques and cost-effectiveness of treatment strategies. In addition, he wrote a thesis on possible improvements of the FDA regulation for new spinal implants.

In the future, he hopes to combine clinical work with his business and entrepreneurial background. In addition, he will continue his research on spinal biomechanics.
Spinal stability following single level lumbar laminectomy: A biomechanical study of the human spine.
Spinal stability following single level lumbar laminectomy

A biomechanical study of the human spine

Arno Bisschop
Thesis: Spinal stability following single level lumbar laminectomy: A biomechanical study of the human spine

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Spinal stability following single level lumbar laminectomy

A biomechanical study of the human spine
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Contents

Chapter 1
General introduction

Chapter 2
The impact of bone mineral density and disc degeneration on shear strength and stiffness of the lumbar spine following laminectomy

Chapter 3
Which factors prognosticate shear instability following lumbar laminectomy?

Chapter 4
Torsion biomechanics of the spine following lumbar laminectomy: a human cadaver study

Chapter 5
Which factors prognosticate rotational instability following lumbar laminectomy?

Chapter 6
Effects of repetitive movement on range of motion and stiffness around the neutral orientation of the human lumbar spine

Chapter 7
Lumbar laminectomy alters the biomechanical behavior of the treated segment without affecting adjacent segments
Clinical Biomechanics (2014) 29: 912 – 917

Chapter 8
The effects of single level instrumented laminectomy on adjacent biomechanics
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Chapter 9
General discussion

Chapter 10
Nederlandse samenvatting

Appendices
List of abbreviations
Contributors
Commission
Acknowledgements
List of publications related to this thesis
List of presentations and awards related to this thesis
About the author
1

General introduction
The degenerative lumbar spine

The human lumbar spine is a complex structure providing a balance between stability and functionality. It supports the human torso in daily activity and protects neural structures, which are closely related to lumbar spinal anatomy. The lumbar spine consists of five vertebral bodies (L1 – L5) that connect the facet joints through their pedicles. The facet joints are, together with the intervertebral discs, the articulating parts of the lumbar spine enabling limited shear, flexion, extension, lateral bending and axial rotation. The whole of bony structures and intervertebral discs is further supported by soft tissue structures including ligaments (Figure 1). Neural structures, including the spinal cord and cauda equina are located in the central canal. In the intervertebral foramina, nerve roots are present.

**Figure 1.** On the left, an untreated human lumbar spine (L1 – L5) is presented. On the right the similar lumbar spine is shown on which a laminectomy on level L4 is performed. In addition, strategies for undercutting are provided with dotted lines on the lower right image. Degenerative changes are not presented in this figure.
The natural process of ageing can lead to degenerative spinal stenosis. Degenerative spinal stenosis is a condition where multiple anatomical pathophysiological mechanisms are involved, all starting with spondylosis, i.e. degeneration of the spinal segments 27. First, the intervertebral disc loses its ability to contain water, which leads to a loss of disc height and subsequent intervertebral disc degeneration As a result of disc degeneration, increasing stress leads to adaptation of bony structures such as the vertebral endplates, development of osteophytes and thickening of the posterior longitudinal and flavum ligament. All these degenerative changes can lead to narrowing of the central spinal canal and foramina in which the nerve roots are located. Clinically, degenerative lumbar spinal stenosis is most seen on level L3 – L4 and L4 – L5 1. Specifically for the lumbar spine, spinal canal stenosis can be caused by the protrusion of the annulus fibrosis because of degeneration, optionally with a bulging disc, and by osteophytes from the posterior vertebral bodies. On the posterior side of the lumbar spine, narrowing can be caused by degenerative changes around the facet joints or facet joint hypertrophy with thickening of the flavum ligament, which can cause neural compression.

Secondly, compression of neural elements due to narrowing of the spinal canal can also be secondary to a degenerative spondylolisthesis. In degenerative lumbar spondylolisthesis the loss of intervertebral disc height, degenerative changes and sagittal orientation of the facet joints cause abnormal mobility 26. As a result, the facet joint surfaces are able to slide over each other, causing a forward shift of one vertebra relative to another, thereby decreasing spinal canal space. Degenerative spondylolisthesis can cause both cauda equina- and root compression. Usually, the upper vertebra slides anteriorly relative to the lower vertebral body. Degenerative lumbar spondylolisthesis most often involves a slip of L4 relative to L5, usually occurs in patients older than 50 years and is significantly more common in women than in men 31. In the occasion that one facet joint slips more than the other, a characteristic rotational slip is seen. A rotational component can lead to a (degenerative) scoliosis or rotoscoliosis 36. Degenerative spondylolisthesis is sometimes associated with the presence of spondylolysis, i.e. a fracture in the pars interarticularis.

Thirdly, degenerative lumbar spinal stenosis can also develop as a result of traumatic spinal fractures or osteoporotic compression fractures leading to subsequent narrowing of the spinal canal.
Verbiest - Historical perspective

Henk Verbiest (1909 - 1997) was the first to describe the relation between neurological symptoms and neural compression due to lumbar spinal stenosis. He started his career as a neurologist-psychiatrist and later specialized in neurosurgery in Paris in 1938. The first time that Verbiest hypothesized the lumbar stenosis syndrome or the Verbiest syndrome dated from the late forties. In 1950, Verbiest’s ideas were published in Het Nederlands Tijdschrift voor de Geneeskunde. It took until 1954 before it was internationally accepted in The Journal of Bone and Joint Surgery. In this paper he presented a case series of seven individual patients on which he performed myelography. Patients with degenerative spinal stenosis showed a block with the appearance of extradural compression. This anatomical anomaly was confirmed during surgery. Verbiest continued his career as a neurosurgeon in the Academic Hospital in Utrecht the Netherlands, where he was appointed as Professor in 1963. He became one of the leading neurosurgeons in the world. In 1981 he retired as a Professor. Verbiest died in 1997 at the age of 88 years old.

Degenerative lumbar spinal stenosis: clinical perspective

Generally, patients suffering from degenerative lumbar spinal stenosis already have a long history of low back pain and nerve root pain that is associated with activity and improve with rest. Low back pain and nerve root pain are often a-specific. Stress inducing moments, i.e. coughing, sneezing and straining rarely aggravate complaints. Besides low back pain, symptoms of degenerative lumbar spinal stenosis can include tingling, numbness and sometimes muscle weakness in one or both legs. Symptoms increase with walking and disappear with rest, preferably in a sitting position. Forward bending often clearly diminishes symptoms. So-called neurogenic claudication can be diagnosed only after vascular claudication is excluded. Pain and paresthesia generally start (uni- or bilaterally) in the thigh and spread through the leg, sometimes with sensory and motor deficiencies. In addition, reduced reflexes may be seen during neurological examination. Walking ability decreases gradually, advancement of symptoms forces the patient to stop walking after short distances. In most cases, cycling remains possible. Eventually a cauda equina syndrome can develop with symptoms as loss of anal sphincter function, incontinence and impotence. A progressive presentation of neurological symptoms often justifies a diagnostic MRI (Figure 2). Standing upright radiographs are considered standard practice and are generally made in an earlier phase of treatment. In case there is a contra indication for an MRI, a CT myelography can be considered.
Figure 2. Typical example of T1-weighted MRI images presenting degenerative lumbar spinal stenosis. Above a sagittal image with evident degenerative changes on levels L4 – L5 and L5 – S1 is shown. Below a transversal coupe of L4 – L5 is shown in which the spinal canal is compromised.
The treatment of spinal stenosis is often conservative, i.e. avoiding painful activities, explanation of the cause of symptoms, prescription of NSAIDs and possibly administering epidural steroid injections. If a well-conducted conservative management fails, surgical decompressive therapy is usually the next step. The aim of a spinal decompression operation is the removal of all compressing tissues, including both bone and soft tissues, with preservation of spinal stability. One of the most commonly used methods to decompress the lumbar spine is by performing a facet-sparing laminectomy. In case a decompressed segment is thought to progress into an unstable segment, i.e. possible progression in spondylolisthesis or spondylolysis, additional stabilization can be added. The clinical success of a decompression ranges from 70 to 80%. Degenerative spondylolisthesis is more likely to develop after a previous uninstrumented decompression procedure such as a laminectomy. In case an iatrogenic spondylolisthesis develops, the results of treatment are less optimal and re-operation might be considered.

In patients over 65 years of age, symptomatic degenerative lumbar spinal stenosis is the most common indication for spinal surgery. Due to the ageing population an increase in absolute numbers of surgical interventions is expected to be present in the next decades, especially since elderly patients demand a higher level of daily life activity nowadays.

**Lumbar laminectomy for degenerative spinal stenosis**

When conservative treatment for degenerative spinal stenosis fails, a surgical intervention is generally considered. Several decompressive techniques, including laminectomy, hemi-laminectomy, laminotomy, foraminotomy and facetectomy have shown to be beneficial in the past. These techniques are all primarily used to decompress trapped nerve roots or cauda equina in order to alleviate symptoms. Among these types of decompression, facet-sparing laminectomy of the lumbar spine is the most frequently used therapy of choice for degenerative lumbar spinal stenosis.

The performance of an operative laminectomy includes removal of the spinous process and part of the lamina while the facet joints are generally left intact (Figure 1). If necessary, undercutting of the inferior facet joint is performed in addition to a laminectomy. By using this technique, the integrity of supraspinous, interspinous and flavum ligaments is lost and all spinal structures attached to the spinal process and part of the lamina are partially or completely disrupted. Hereby, facet joints should remain intact because removal increases the risk of instability. Following laminectomy, impinged nerves are decompressed in an effort to relieve neurological symptoms, such as leg pain, tingling, numbness and weakness.
are relieved. In addition, it has been shown that quality of life (QoL) improves after decompressive laminectomy.

Although laminectomy resolves complaints in most cases, the anatomical alterations caused by this technique might affect lumbar spinal biomechanics. As a result, a laminectomy can, besides the risk of developing an epidural hematoma and infection, lead to complications such as symptomatic postoperative lumbar instability or even postoperative failure of the treated spinal motion segment. This clinical situation is often called a post-laminectomy syndrome. When this syndrome develops, or is expected to develop due to an extensive per-operative bony removal, additional instrumentation is applied in order to stabilize the decompressed lumbar spinal segment. In clinical practice, placement of a posterior pedicle screw and rod construction with or without the use of bone graft to pursue spinal fusion is chosen. Another option to treat degenerative lumbar spinal stenosis is by using interspinous devices. However, these implants were found to have a high reoperation rate and are not frequently used anymore.

In order to improve clinical practice, it is of paramount importance to investigate spinal biomechanics following laminectomy. Knowledge concerning the need for, and also about potential negative effects of, additional instrumentation is required. A main aim of this thesis is therefore first to provide information that allows for improved prediction of mechanical alterations after laminectomy, and second to assess effects of additional instrumentation. Thereby, not only the treated segment but also the adjacent segments will be considered. In all cases, a biomechanical approach is chosen. Before outlining this thesis, first a summary of current scientific knowledge regarding clinical and biomechanical effects of laminectomy and instrumentation is provided.

**Spinal instability**

Instability of a spinal segment is an important concept in this thesis. From a biomechanical point of view, White and Panjabi were one of the first to describe spinal instability. These authors proposed a definition of clinical spinal instability as the loss of the spine to limit its movements under physiological loads such that neurological disturbances (e.g. through nerve root compression), deformation (e.g. scoliosis, spondylolisthesis), or pain are prevented. However, Reeves et al. previously postulated that the term stability is often used ambiguously in spinal biomechanics. Therefore, to be able to frame our understanding of instability, we defined spinal instability from a biomechanical perspective as a significant and substantial increase in range of motion (ROM), and/or reduced stiffness (around the neutral
zone in submaximal testing experiments) and/or a decrease in ultimate spinal strength \(^{45, 46}\). In addition, we add yielding to these commonly used biomechanical parameters in our biomechanical evaluations. The yield point represents the transition from elastic to plastic deformation. The yield point might be one of the most critical biomechanical properties because it marks the beginning of the irreversible deformation of a spinal motion segment \(^{52}\).

Laminectomy and instrumentation: clinical effects on the treated segment

Postoperative segmental instability of the lumbar spine, i.e., postoperative iatrogenic spondylolisthesis is a frequently reported complication reported following laminectomy \(^{30}\). Symptomatic post-operative instability justifies re–operation to stabilize and fuse the unstable segment \(^{32}\). The incidence of post–facet sparing laminectomy spondylolisthesis has been reported to range from 8 % – 31 % \(^{23, 25}\).

Although facet joints are often preserved during a laminectomy, the lamina, interspinous-, supraspinous-, and flavum ligaments are incised and/or removed. These structures all have a stabilizing function for a spinal motion segment. Removal of these structures might cause clinically relevant spinal instability, defined as an increase in range of motion (ROM), reduced stiffness and/or a decrease in ultimate spinal strength of the spine that leads under physiological loads to neurological disturbances (e.g. through nerve root compression), deformation (e.g. scoliosis, spondylolisthesis), and / or pain \(^{67}\). Remarkably however, biomechanical effects of this type of single level decompression have not been thoroughly investigated. Furthermore, it is unknown whether these effects can be predicted in order to improve clinical practice by pre–operative identification of segments that have a high probability of progression into an unstable clinical segment. If this would be possible, surgeons could predict which patients might benefit from additional instrumentation to stabilize the involved segment. At present, there are no strict guidelines on when additional instrumentation should be added.

Currently, during surgical planning and in the operation theatre the expert opinion of the surgeon considering the use of instrumentation in addition to a decompressive laminectomy is decisive. In case additional instrumentation techniques are used, mostly, a posterior pedicle screw and rod construction is chosen with additional bone graft, which fuses the spinal segment. Posterior fusion has been acknowledged as the standard surgical practice in the decompressed lumbar spine that might evolve to a clinical unstable segment \(^{16, 22, 30}\). With the use of posterior instrumentation, a surgeon also strives for arthrodesis, i.e. fusion of a segment by bone formation \(^{8}\). Successful arthrodesis of a segment after posterior lumbar fusion is
seen in majority of cases and provides good clinical results 24, 34, 37. However, the procedure of a posterior spondylodesis itself increases the probability of additional implant-related complications, including infection, neurological impairment, adjacent segment disease (ASD), increased blood loss during surgery, extended surgery time, and instrumentation failure 16, 18. The probability of implant-related complications therefore needs to be weighted against the risk of postoperative complications after laminectomy without stabilization 43. Moreover, it has been stated that pedicle screws do not necessarily increase the stability of a lumbar spinal segment since pull-out strength is limited by a patient’s bone quality 60. Especially in the elderly female patient, bone mineral density (BMD) might be a limiting factor. Finally, it should also be noted that the use of instrumentation also significantly increases costs of surgery 16.

**Laminectomy and instrumentation: clinical effects on the adjacent segments**

The possibility of adjacent segment disease, leading to adjacent spondylosis in the long term, is receiving increasing interest among spine surgeons. It is progressively becoming clear that, in particular, the use of posterior lumbar interbody fusion (PLIF) and posterior lumbar fusion (PLF) hold a risk of ASD 50. Pathology is often found at the level proximal to spinal fusion 10. ASD is one of the reasons why re-operations are necessary and has a higher incidence in elderly patients 38. The overall probability of undergoing revision surgery for ASD was 5.8 % at five years and 10.4 % and ten years postoperative. The prevalence of ASD requiring re-operation in patients older than 60 years of age was 21.9 % at ten years postoperative.

The exact pathogenesis of ASD is not well understood. First, it has been argued that ASD is mere the development of physiological degeneration. Secondly, since spinal biomechanics may change at the adjacent level after application of instrumentation, it is argued that a spondylodesis causes a progress of this particular natural path of degeneration. The first argument does not explain why only the adjacent segment progresses into a state of degeneration. The second argument is not a logical explanation from a biomechanical perspective since after a spondylodesis, there is no reason why moments and forces in adjacent segments would change more than the other segments cranial to a spondylodesis. Furthermore, it does not explain the clinical finding of ASD in proximal and not in distal segments relative to the treated segment. Purely pragmatic, typically topping-off procedures are used to prevent or slow down ASD 42. Topping–off procedures combine rigid fusion with a flexible pedicle screw system to prevent ASD 56. Another factor associated with the progress of ASD is sagittal imbalance 50. The biomechanical effects of a single level lumbar
laminectomy, as well as of instrumented laminectomy on the adjacent segments are unknown yet. Obviously, clinical practice is better substantiated when adequate information on the biomechanical effects of laminectomy and subsequent instrumentation on the treated and adjacent segments is available.

Biomechanics of the lumbar spine

In daily life, the lumbar spine is subjected to compression and shear forces, torsion moments and loads in flexion and extension and lateral bending. Magnitudes of these forces differ substantially between persons due to whether or not physically demanding tasks are performed and on the basis of body weight. While compression forces generally stiffen the spine, all other loads may be relevant with regard to spinal instability as defined in this thesis.

It is commonly held that shear forces are the main cause of spondylolisthesis. Shear forces over 1000 N at the lower lumbar levels commonly occur in the lumbar spine. In vitro experiments have shown that (fatigue) shear loading is bared by both the intervertebral disc and neural arch. In the initial shear loading phase, particularly the neural arch is susceptible to these forces because it is much stiffer than the intervertebral disc. This could explain why fracturing of the pars interarticularis or spondylolysis is frequently reported. Therefore, it is quite likely that decreased resistance to shear loading after laminectomy plays a crucial role in the incidence of post-operative complications, or post-laminectomy syndrome. Indeed, both in vitro and finite element studies have shown that the shear stiffness of the spinal segment after laminectomy, both combined and not combined with other decompression techniques, is reduced. In addition, laminectomy as well as removal of posterior elements substantially reduces spinal shear strength. However, the latter has only been shown for porcine spines. Effects of laminectomy on shear biomechanics of the human lumbar spine are unknown at present.

Not only shear forces can be a cause of the post-laminectomy syndrome, also torsional injuries of the lumbar spine can occur when the applied load includes substantial axial rotation (AR). Such damage may, at a later stage, also lead to symptomatic spondylolisthesis and/or degenerative scoliosis. During daily activities such as asymmetric lifting, the lumbar spine is subjected to torsion moments, resulting in axial rotation at the lumbar levels. It is held that a decreased resistance to spinal torsion is one of the most important parameters in the etiology of low back pain and disc degeneration. Therefore, it might be argued that laminectomy also makes the lumbar spine more susceptible to rotational injury. A reduction in rotational stability after laminectomy is likely since posterior elements, consisting of both
soft and bony tissue, are crucial in restraining axial rotation. Torsional strength of the untreated lumbar spine was studied in the past. The effects of laminectomy on torsion biomechanics are, however, unknown.

During daily loading, the spine is not only subjected to shear and torsion loads but also to submaximal loading in flexion and extension (FE) and lateral bending (LB). Spinal FE, LB and AR before and after surgical interventions are often evaluated in submaximal loading devices. Considering decompressive surgery, it has been shown in the past that extensive decompressive techniques destabilize the whole lumbar spine, i.e. a substantial increase in ROM and a decrease in stiffness in all three motion directions. This destabilization could lead to altered motion patterns and may therefore, again, increase the risk of compression of neurological structures and injury including structural damage of the intervertebral disc and bony structures. The effects of single level lumbar laminectomy, however, have not yet been evaluated in such a manner. Furthermore, single level lumbar laminectomy might also alter adjacent level biomechanics. Only a limited number of studies investigated biomechanical behaviour of segments adjacent to decompressive surgery. Outcomes of these studies vary. Instability at adjacent levels has clinical relevance since these levels might develop ASD. Little is known about the adjacent segment biomechanics after laminectomy with or without the addition of instrumentation.

In order to investigate the above-presented questions and ambiguities considering single level lumbar laminectomy, a series of biomechanical experiments will be presented in this thesis, including testing procedures until failure and submaximal experiments. All biomechanical experiments are conducted on elderly fresh frozen human cadaveric lumbar spines in order to mimic clinical conditions as accurately as possible.

**Aims of the present thesis: unsolved biomechanical issues**

Based on the considerations given above, this thesis addresses three major questions. *First.* How does single level lumbar laminectomy affect spinal ROM, stiffness and strength? *Secondly.* Can spinal instability following lumbar laminectomy be prognosticated based on clinically measurable imaging parameters? *Thirdly.* What are the biomechanical effects under submaximal loading conditions of single level laminectomy on the treated and adjacent levels with and without instrumentation of the treated segment?
The specific aims of the subsequent chapters of this thesis are:

1. To assess the effects of single level lumbar laminectomy on shear biomechanics of the treated segment (Chapter 2).
2. To assess how shear biomechanics after single level lumbar laminectomy can be predicted (Chapter 3).
3. To assess the effects of single level lumbar laminectomy on torsion biomechanics of the treated segment (Chapter 4).
4. To assess how torsion biomechanics after single level lumbar laminectomy can be predicted (Chapter 5).
5. To assess the methodology for submaximal biomechanical testing of lumbar spinal segments (Chapter 6).
6. To quantify the effects of laminectomy on submaximal biomechanical behaviour of treated and adjacent lumbar segments (Chapter 7).
7. To quantify the effects of instrumented laminectomy on submaximal biomechanical behaviour of treated and adjacent lumbar segments (Chapter 8).

In the General discussion (Chapter 9), the aim is to provide a flow-chart based on the results of the preceding chapters in order to advise in surgical decision-making during decompressive lumbar laminectomy.
References


The impact of bone mineral density and disc degeneration on shear strength and stiffness of the lumbar spine following laminectomy


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Abstract

Purpose
Laminectomy is a standard surgical procedure for elderly patients with symptomatic degenerative lumbar stenosis. The procedure aims at decompression of the affected nerves, but it also causes a reduction of spinal shear strength and shear stiffness. The magnitude of this reduction and the influence of bone mineral density (BMD) and disc degeneration are unknown. We studied the influence of laminectomy, BMD, and disc degeneration on shear force to failure (SFF) and shear stiffness (SS).

Methods
Ten human cadaveric lumbar spines were obtained (mean age 72.1 years, range 53 – 89 years). Laminectomy was performed either on L2 or L4, equally divided within the group of ten spines. BMD was assessed by dual X-ray absorptiometry (DXA). Low BMD was defined as a BMD value below the median. Intervertebral discs were assessed for degeneration by MRI (Pfirrmann) and scaled in mild and severe degeneration groups. Motion segments L2 – L3 and L4 – L5 were isolated from each spine. SFF and SS were measured, while loading simultaneously with 1600 N axial compression.

Results
Low BMD had a significant negative effect on SFF. In addition, a significant interaction between low BMD and laminectomy was found. In the high BMD group, SFF was 2482 N (range 1678 – 3284 N) and decreased to 1371 N (range 940 – 1886 N) after laminectomy. In the low BMD group, SFF was 1339 N (range 909 – 1628 N) and decreased to 761 N (range 561 – 1221 N). Disc degeneration did not affect SFF, nor did it interact with laminectomy. Neither low BMD nor the interaction of low BMD and laminectomy did affect SS. Degeneration and its interaction with laminectomy did not significantly affect SS.

Conclusions
In conclusion, low BMD significantly decreased SFF before and after lumbar laminectomy. Therefore, DXA assessment may be an important asset to preoperative screening. Lumbar disc degeneration did not affect shear properties of lumbar segments before or after laminectomy.
**Introduction**

Symptomatic degenerative spinal stenosis is the most common indication for lumbar surgery in patients over 65 years of age. Symptoms include leg, back and buttock pain, radiculopathy, neurogenic claudication and subjective muscle weakness. The standard surgical procedure for symptomatic degenerative lumbar spinal stenosis is decompression of the spinal canal by laminectomy. Despite preservation of the facet joints, the anatomical integrity and stabilizing function of bony structures, and interspinous, supraspinous and flavum ligaments is lost. Most complications reported after laminectomy are related to postoperative segmental instability of the lumbar spine or postoperative spondylolisthesis. The increased occurrence of postoperative fractures of the pars interarticularis following laminectomy might result from a reduced resistance to shear forces.

Both in vitro and finite element studies have shown that the shear stiffness of the spinal segment after laminectomy, whether or not combined with other decompression techniques, is reduced. In addition, laminectomy as well as removal of posterior elements substantially reduces spinal strength in shear loading. In vitro experiments have also shown that shear loading can lead to bony failure of the posterior elements of the spine, with fracturing of the pars interarticularis most often reported. Therefore, it is quite likely that decreased resistance to shear loading after laminectomy plays a crucial role in the incidence of postoperative complications, i.e. the so-called post-laminectomy syndrome.

The relation between laminectomy combined with partial facetectomy and the reduction of shear force to failure (SFF) has recently been studied in healthy young porcine spines. The relative SFF reduction was quite limited (22%). The shear stiffness (SS) of these spinal segments was reduced by 9% after treatment.

However, it has been shown that a normal, healthy disc substantially contributes to shear resistance. In contrast to young porcine spines, the spines of patients with degenerative lumbar stenosis often show signs of intervertebral disc degeneration. In addition, it has been shown that degenerative changes in the intervertebral disc affect the mechanical properties of the lumbar motion segment. Yet, the interaction between the effects of laminectomy and disc degeneration of the lumbar spine has not been studied. Furthermore, it is well known that low bone mass affects the risk of vertebral fracture. Hence, low bone mineral density (BMD) potentially could reduce SFF and SS and may interact with laminectomy as well.

In the present study, we quantified the effects of laminectomy on the SFF and SS in ten human cadaveric lumbar spines. We also assessed the relation of these biomechanical features...
with BMD and disc degeneration. We hypothesized that laminectomy substantially reduces SFF and SS of the human lumbar spine, and that BMD and disc degeneration influence the residual SFF and SS after laminectomy.

**Methods**

**Specimens and specimen preparation**

Thoracolumbar spines (T12 – L5) were harvested from freshly frozen (− 20 °C) human cadavers (mean age 72.1 years, range 53 – 89 years). None of the deceased subjects had any history of spinal injury, spinal surgery or spinal metastatic disease. The spines were thawed before testing. Excessive soft tissue and muscle tissue were carefully removed, keeping the anterior and posterior longitudinal ligaments as well as the facet joints intact. Laminectomy was performed at level L2 of five randomly chosen thoracolumbar spines, and at level L4 of the remaining five thoracolumbar spines. The untreated level of each thoracolumbar spine was considered as internal control. Since correlations between mechanical properties of segments from the same spine are considered high, the adjacent segment from the same spine was used as control. To exclude systematic effects of segment level, laminectomy was performed randomly at L2 or L4. Laminectomy, analogous to standard clinical practice, was performed by removing the spinous process and part of the lamina. During preparation, assessment and biomechanical testing, specimens were kept hydrated using 0.9 % saline-soaked gauzes. Furthermore, anteroposterior, lateral and oblique radiographs were made to determine whether bridging osteophytes were present in segments. Thoracolumbar spines with bridging osteophytes were excluded from this study.

Before testing, BMD (g/cm²) of each lumbar spinal section (L1 – L4) was measured with dual X-ray absorptiometry (DXA, Hologic QDR 4500 Delphi DXA scanner, Waltham, MA, USA) in anteroposterior direction. Low BMD was defined as lower than median, while high BMD was defined as median or higher. Dichotomized BMD was related as an independent variable to biomechanical outcomes in a generalized estimating equations (GEE) model.

MRI (Siemens Symphony 1.5 T: Syngo MR A30, software NUMARIS/4, Berlin, Germany) of T12 – L5 segments was performed to assess disc degeneration. Degeneration of intervertebral discs was graded according to the Pfirrmann classification for lumbar spinal degeneration. Grading was performed on T2-weighted mid-sagittal sections. Subsequently, degeneration scores were dichotomized; grades 3 or lower were classified as ‘mildly degenerated’, while grades higher than 3 were classified as ‘severely degenerated’. Dichotomized disc degeneration was related as an independent variable to biomechanical outcomes in a GEE model.
Subsequently, each spinal segment (T12 – L5) was dissected into two motion segments (L2 – L3 and L4 – L5). The motion segments were potted in a casting mold using low melting point (48 °C) bismuth alloy (Cerrolow – 147; 48 % bismuth, 25.6 % lead, 12 % tin, 9.6 % cadmium, and 4 % indium). The disc was placed parallel based on visual inspection to the flat surface of the bismuth. The upper and lower vertebral bodies were fixed securely into the alloy by adding screws into the vertebral body. Screw fixation was reinforced with orthopedic bone cement (Stryker, Simplex, Kalamazoo, MI, USA). All articulating parts were kept free.

**Biomechanical testing procedure**

The casting mold was placed in a hydraulic materials testing machine (Instron, model 8872; Instron and IST, Norwood, Canada) (Figure 1). The caudal vertebral body was fixed on a plateau that allowed movement in axial and transverse directions only. Transverse movements were allowed, so segments were able to find their physiological motion patterns and to correct for possible differences in embedding. Segments were loaded with an axial compressive force of 1600 N \(^{19}\). A pure axial compressive force was applied using a pneumatic cylinder. Calibration of axial compression was performed using a load cell (Hottinger Baldwin Messtechnik, Force Transducer Type C2, Darmstadt, Germany). Since compression was applied in a purely axial direction, bending moments were minimized. The chosen amount of preload was selected to allow for comparison with previous work \(^{19}\). Subsequently, anterior shear load was applied with a constant rate of 2.0 mm/min on the casting mold containing the cranial vertebral body, until failure of the vertebral motion segment \(^{20}\). The test was stopped after hearing a clear crack or after a large force reduction was seen. Shear force and displacement were digitized and stored at 100 samples per second (Instron Fast Track 2, Norwood, Canada).

For each of the 20 motion segments tested, SFF and displacement at the instant of failure (DF) were determined. Failure was defined as the point at which maximum load was recorded in the load–displacement curves (Figure 2) for each specimen. The SFF was defined as the maximum force in Newton until failure. The average SS was calculated from the load–displacement curve. SS was calculated between 25 % and 75 % of the SFF. SS was estimated by means of a least squares fit of a straight line through the shear force–displacement data with the slope of the fitted line representing stiffness. The deformation in this region was linear, with an \(r^2 > 0.925\) (Table 1) between load and displacement for all motion segments.
Figure 1. Segment placed in the materials testing machine, showing the fixed center of rotation (A), the free center of rotation (B) and finally vertical load transfer through to apply shear loads (C).
Figure 2. Load–displacement curve of two motion segments from one human cadaver. In this case segment L2 – L3 was with laminectomy, while L4 – L5 was untreated.

Statistical methods

Generalized estimating equations (GEE) were used to assess relationships between dependent and independent variables. The GEE is a regression analysis that takes the repeated measures character of the data into account. Dependent variables were SFF, DF and SS. First, analyses were performed to determine the effect of laminectomy on all three dependent variables with correction for the confounding effect of segment level. Segment level was added to the GEE model as a dichotomous independent variable. Next, we tested whether dichotomized BMD and disc degeneration co-determined independent variables and whether these modified the effects of laminectomy, by adding an interaction term to the model, which was omitted when not significant. The same procedure was used for dichotomized Pfirrmann scores. A significance level of 5 % was used. The statistical analyses were performed using SPSS for Mac version 16.0 (SPSS Incorporated, Chicago, IL, USA).
Table 1. Specimens, independent and dependent variables per segment. For shear stiffness, $r^2$ values are added in brackets.

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Segment</th>
<th>Laminectomy</th>
<th>Total bone mineral density of L1-L4 (BMD) (g/cm²)</th>
<th>Disc degeneration (Pfirrmann) (1-5)</th>
<th>Shear force to failure (SFF) (N)</th>
<th>Displacement at failure (DF) (mm)</th>
<th>Shear stiffness (SS) (N/mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Specimen 01 (Male, 79)</td>
<td>L2-L3</td>
<td>0</td>
<td>0.87</td>
<td>4</td>
<td>2317</td>
<td>11.2</td>
<td>282 (0.994)</td>
</tr>
<tr>
<td></td>
<td>L4-L5</td>
<td>1</td>
<td>3</td>
<td>1258</td>
<td>7.9</td>
<td>164 (0.997)</td>
<td></td>
</tr>
<tr>
<td>Specimen 02 (Male, 53)</td>
<td>L2-L3</td>
<td>0</td>
<td>0.89</td>
<td>2</td>
<td>3284</td>
<td>15.9</td>
<td>188 (0.998)</td>
</tr>
<tr>
<td></td>
<td>L4-L5</td>
<td>1</td>
<td>3</td>
<td>1886</td>
<td>9.4</td>
<td>234 (0.997)</td>
<td></td>
</tr>
<tr>
<td>Specimen 03 (Male 72)</td>
<td>L2-L3</td>
<td>0</td>
<td>0.95</td>
<td>5</td>
<td>1678</td>
<td>9.3</td>
<td>147 (0.925)</td>
</tr>
<tr>
<td></td>
<td>L4-L5</td>
<td>1</td>
<td>5</td>
<td>1775</td>
<td>8.6</td>
<td>270 (0.996)</td>
<td></td>
</tr>
<tr>
<td>Specimen 04 (Female, 82)</td>
<td>L2-L3</td>
<td>0</td>
<td>0.45</td>
<td>4</td>
<td>909</td>
<td>3.9</td>
<td>238 (0.995)</td>
</tr>
<tr>
<td></td>
<td>L4-L5</td>
<td>1</td>
<td>5</td>
<td>561</td>
<td>2.1</td>
<td>341 (1.000)</td>
<td></td>
</tr>
<tr>
<td>Specimen 05 (Male, 78)</td>
<td>L2-L3</td>
<td>0</td>
<td>0.51</td>
<td>4</td>
<td>1292</td>
<td>6.8</td>
<td>275 (0.998)</td>
</tr>
<tr>
<td></td>
<td>L4-L5</td>
<td>1</td>
<td>4</td>
<td>1221</td>
<td>7.6</td>
<td>212 (0.998)</td>
<td></td>
</tr>
<tr>
<td>Specimen 06 (Male, 79)</td>
<td>L2-L3</td>
<td>1</td>
<td>0.65</td>
<td>2</td>
<td>994</td>
<td>6.3</td>
<td>152 (0.997)</td>
</tr>
<tr>
<td></td>
<td>L4-L5</td>
<td>0</td>
<td>3</td>
<td>2408</td>
<td>9.9</td>
<td>323 (0.996)</td>
<td></td>
</tr>
<tr>
<td>Specimen 07 (Male, 62)</td>
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<td>1</td>
<td>0.91</td>
<td>2</td>
<td>940</td>
<td>5.7</td>
<td>191 (0.998)</td>
</tr>
<tr>
<td></td>
<td>L4-L5</td>
<td>0</td>
<td>5</td>
<td>2724</td>
<td>17.7</td>
<td>254 (0.998)</td>
<td></td>
</tr>
<tr>
<td>Specimen 08 (Female, 64)</td>
<td>L2-L3</td>
<td>1</td>
<td>0.58</td>
<td>3</td>
<td>660</td>
<td>3.4</td>
<td>178 (0.993)</td>
</tr>
<tr>
<td></td>
<td>L4-L5</td>
<td>0</td>
<td>3</td>
<td>1553</td>
<td>7.5</td>
<td>214 (0.968)</td>
<td></td>
</tr>
<tr>
<td>Specimen 09 (Female, 63)</td>
<td>L2-L3</td>
<td>1</td>
<td>0.57</td>
<td>3</td>
<td>641</td>
<td>5.7</td>
<td>83 (0.973)</td>
</tr>
<tr>
<td></td>
<td>L4-L5</td>
<td>0</td>
<td>3</td>
<td>1313</td>
<td>5.0</td>
<td>332 (0.998)</td>
<td></td>
</tr>
<tr>
<td>Specimen 10 (Female, 89)</td>
<td>L2-L3</td>
<td>1</td>
<td>0.59</td>
<td>4</td>
<td>721</td>
<td>3.9</td>
<td>192 (0.994)</td>
</tr>
<tr>
<td></td>
<td>L4-L5</td>
<td>0</td>
<td>4</td>
<td>1628</td>
<td>11.6</td>
<td>263 (0.994)</td>
<td></td>
</tr>
</tbody>
</table>

0: untreated; 1: laminectomy
Results

All specimen parameters and outcome measures are presented in Table 1. MRI images and visual inspection confirmed that facet joints were intact and no fractures of the pars interarticularis were present in operated or intact segments before mechanical testing.

The median total BMD of all sections (L1 – L4) was 0.62 g/cm² (range 0.45 – 0.95 g/cm²). Therefore, low BMD was defined as < 0.62 g/cm² and high BMD as ≥ 0.62 g/cm². Ten segments were classified, according to Pfirrmann 14, as mildly degenerated and ten segments as severely degenerated.

All statistical results are presented in Table 2. Laminectomy resulted in a decrease of SFF (44.2 %), DF (38.6 %) and SS (19.9 %), which was significant according to the GEE models with laminectomy and level as independent variables.

Low BMD had a significant negative effect on SFF. In addition, a significant interaction between low BMD and laminectomy was found (Figure 3 & Table 2). In the high BMD group, SFF was 2482 N (range 1678 – 3284 N) and decreased to 1371 N (range 940 – 1886 N) after laminectomy. In the low BMD group, SFF was 1339 N (range 909 – 1628 N) and decreased to 761 N (range 561 – 1221 N). Disc degeneration based on a Pfirrmann scale did not affect SFF, nor did it interact with laminectomy (Figure 4 & Table 2). Low BMD also reduced DF. The interaction between low BMD and laminectomy did not reach significance (Figure 3 & Table 2). Disc degeneration was not found to affect DF (Figure 4 & Table 2).

Neither low BMD nor the interaction of low BMD and laminectomy did affect SS (Figure 3 & Table 2). Degeneration and its interaction with laminectomy did not significantly affect SS (Figure 4 & Table 2).
Table 2. Regression coefficients (corrected for segment level) and $P$-values (in brackets with significant values in bold) for the effects of laminectomy, as well as the effects of bone mineral density (BMD) and disc degeneration and their interactions with laminectomy on shear force to failure (SFF), displacement at failure (DF), and shear stiffness (SS). Each row in the table represents a regression equation. For example, the second row should be read as: $SFF = 2375 \pm 1271$ (laminectomy: 0 or 1) – 1197 (low BMD: 0 or 1) + 676 (laminectomy: 0 or 1 x low BMD: 0 or 1).

<table>
<thead>
<tr>
<th></th>
<th>Intercept</th>
<th>Laminectomy</th>
<th>Bone mineral density (BMD)</th>
<th>Bone mineral density (BMD)</th>
<th>Disc degeneration (Pfirrmann)</th>
<th>Disc degeneration (Pfirrmann)</th>
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</thead>
<tbody>
<tr>
<td>SFF (N)</td>
<td>1895</td>
<td>-1105</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>(&lt; 0.001)</td>
<td>(0.004)</td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
<td>(0.001)</td>
<td>(0.001)</td>
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<td>SFF (N)</td>
<td>2375</td>
<td>-1271</td>
<td>-1197</td>
<td>676</td>
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<td>-</td>
</tr>
<tr>
<td></td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>SFF (N)</td>
<td>1569</td>
<td>-984</td>
<td>-</td>
<td>-</td>
<td>409</td>
<td>622</td>
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<td></td>
<td>(0.008)</td>
<td>(0.114)</td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
<td>(0.138)</td>
<td>(0.051)</td>
</tr>
<tr>
<td>DF (mm)</td>
<td>9.4</td>
<td>-4.4</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>(&lt; 0.001)</td>
<td>(0.019)</td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
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</tr>
<tr>
<td>DF (mm)</td>
<td>11.9</td>
<td>-5.3</td>
<td>-6.3</td>
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<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
<td>(0.071)</td>
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<td></td>
</tr>
<tr>
<td>DF (mm)</td>
<td>5.3</td>
<td>-1.3</td>
<td>-</td>
<td>-</td>
<td>5.1</td>
<td>-0.6</td>
</tr>
<tr>
<td></td>
<td>(&lt; 0.001)</td>
<td>(0.781)</td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
<td>(0.081)</td>
<td>(0.843)</td>
</tr>
<tr>
<td>SS (N/mm)</td>
<td>226</td>
<td>-67</td>
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<td>-</td>
</tr>
<tr>
<td></td>
<td>(&lt; 0.001)</td>
<td>(0.022)</td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
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<td></td>
</tr>
<tr>
<td>SS (N/mm)</td>
<td>219</td>
<td>-71</td>
<td>16</td>
<td>0</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>(0.082)</td>
<td>(0.565)</td>
<td>(0.996)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SS (N/mm)</td>
<td>225</td>
<td>-77</td>
<td>-</td>
<td>-</td>
<td>2</td>
<td>55</td>
</tr>
<tr>
<td></td>
<td>(&lt; 0.001)</td>
<td>(0.040)</td>
<td>(&lt; 0.001)</td>
<td>(&lt; 0.001)</td>
<td>(0.950)</td>
<td>(0.132)</td>
</tr>
</tbody>
</table>
Figure 3. Shear stiffness (SS), displacement at failure (DF) and shear force to failure (SFF) in relation to BMD and laminectomy (mean values ± SD), according to GEE modeling.
Figure 4. Shear stiffness (SS), displacement at failure (DF) and shear force to failure (SFF) in relation to disc degeneration and laminectomy (mean values ± SD), according to GEE modeling.
Discussion

In this study, we investigated the effects of laminectomy, BMD, and disc degeneration and their interaction on SFF and shear stiffness (SS). The results corroborated our hypothesis that SFF and SS after laminectomy are reduced. In addition, we found that BMD did affect SFF but not SS. In contrast, disc degeneration did not act as a determinant of SFF or SS before or after laminectomy.

Laminectomy was found to reduce SFF of the lumbar spine by approximately 50%. A low BMD also reduced SFF by half, relative to a high BMD. In addition, we found an interaction between BMD and laminectomy. As the regression coefficients in Table 2 indicate, this interaction implied a smaller absolute, but similar relative effect of laminectomy in segments with low BMD as in segments with a high BMD. In both, SFF was roughly halved by laminectomy, i.e. from approximately 2400 to 1100 N in segments with high BMD and from approximately 1200 to 500 N in segments with low BMD (Figure 4).

Our findings are in line with results reported in other in vitro and finite element studies, which also showed reduced shear strength and stiffness of the spine after laminectomy. However, declines found in the present study (44.2 % for SFF and 19.9 % for SS) were much larger than those found in a similar study using porcine specimens (22 % for SFF and 9 % for SS), even though facetectomy was performed in addition to laminectomy in the latter study. With facetectomy, a larger reduction of strength and stiffness could be expected.

Differences in results may be due to differences in species, geometry, age, and degeneration. We hypothesized that especially the latter would affect outcomes. Previous studies observed that the intervertebral disc is responsible for 63 – 74 % of the ultimate shear strength. Therefore, the quality of the intervertebral disc was thought to be of great influence. However, differences in disc degeneration within the present sample did not affect SFF and SS of the segments. In this study, disc degeneration was included as a dichotomized variable. The present specimens all showed signs of degeneration, possibly resulting in a too small variance in disc degeneration to detect its effect. Due to the limited availability of donor material, the sample size in this study was small. The effect of degeneration on dependent variables may exist but is likely to be relatively small in the range studied (all degenerated). In addition, the studied range of spinal segments is comparable to the range that is clinically relevant. Moreover, in contrast to our expectations degeneration appeared to enhance SFF and SS rather than to reduce it.
Since the effect of laminectomy on ultimate shear strength and shear stiffness is considerable, chances of developing postoperative pars interarticularis fractures and spondylolisthesis after laminectomy seem quite substantial, especially when the patient has low BMD. It may be questioned whether patients can safely perform physically demanding tasks after lumbar laminectomy. In physically demanding tasks such as lifting, lumbar spine shear forces have been estimated to range from 1000 to 1770 N at the L5 – S1 joint level\(^9\),\(^10\). Therefore, our data (mean SFF after laminectomy was 1066 N) suggest that laminectomy puts a patient at risk of developing a post-laminectomy syndrome when performing demanding activities. The decrease in amplitude of displacement at failure after laminectomy also shows that less ‘slipping’ (i.e. absolute shear displacement) is necessary before trauma occurs.

Standard pre-operative assessment for laminectomy does not include DXA measurement. Our results suggest that knowledge of BMD may be valuable in pre-operative assessment of patients undergoing decompressive surgery. Subjects with low BMD may require additional posterior instrumented stabilization to prevent postoperative instability. In clinical practice, laminectomy is often, but not always, combined with posterior instrumentation and fusion to prevent complications such as spondylolisthesis\(^2\),\(^5\),\(^7\).

The procedure of stabilization itself increases the probability of implant-related complications, including infection, nerve injury, possible adjacent disc degeneration, increased blood loss, extended surgery time, and instrumentation failure\(^3\),\(^4\). Moreover, it significantly increases the costs of surgery\(^3\). The probability of implant-related complications needs to be weighted against the risk of postoperative complications after laminectomy without stabilization\(^12\).

It has been shown that axial compression and shear forces are quite strongly correlated across a range of daily activities\(^18\) and axial compression is known to increase the shear stiffness of the intervertebral disc\(^6\),\(^16\). Therefore, segments were pre-loaded in compression. We selected a compression load of 1600 N because it is physiologically relevant with the applied shear loads, and allows for comparison with previous work\(^19\).

In the present study, DXA measurement was performed after laminectomy for practical reasons. This leads to a slight underestimation of BMD. However, BMD was measured over the complete lumbar section of L1 – L4, and laminectomy was performed on one segment in each spinal segment. Therefore, it is expected that this did not influence the outcomes.

Finally, we used a single loading cycle. Cyclic loading might, through visco-elastic behavior of the intervertebral disc, shift load to the posterior elements\(^19\), thereby possibly enhancing the effect of laminectomy on SFF.
In conclusion, a 44.2% reduction of SFF and a 19.9% reduction in SS due to laminectomy were observed in ten fresh frozen elderly human lumbar spines. Low BMD had a significant negative effect on SFF. In addition, a significant interaction between low BMD and laminectomy was found. Disc degeneration did not affect SFF, nor did it interact with laminectomy. Neither low BMD nor the interaction of low BMD and laminectomy did affect SS. Disc degeneration and its interaction with laminectomy did not significantly affect SS.
References

5. Fox MW, Onofrio BM (1997) Indications for fusion following decompression for lumbar spinal stenosis. Neurosurg Focus 3:e2
Which factors prognosticate shear instability following lumbar laminectomy?


Awarded with the annual Dutch Spine Society (DSS) Award 2012

Arno Bisschop
Barend J. van Royen
Margriet G. Mullender
Cornelis P. L. Paul
Idsart Kingma
Timothy U. Jiya
Albert J. van der Veen
Jaap H. van Dieën
Abstract

Purpose
Reduced strength and shear stiffness (SS) of lumbar motion segments following laminectomy may lead to instability. The purpose of the present study was to assess a broad range of parameters as potential predictors of shear biomechanical properties of the lumbar spine.

Methods
Radiographs and MRI of all lumbar spines were obtained to classify geometry and degeneration of the motion segments. Additionally, dual X-ray absorptiometry (DXA) scans were performed to measure bone mineral content and density (BMC and BMD). Facet sparing lumbar laminectomy was performed either on L2 or L4, in 10 human cadaveric lumbar spines (mean age 72.1 years, range 53 – 89 years). Spinal motion segments were dissected (L2 – L3 and L4 – L5) and tested in shear, under simultaneously loading with 1600 N axial compression. Shear stiffness, shear yield force (SYF) and shear force to failure (SFF) were determined and statistical correlations with all parameters were established.

Results
Following laminectomy, SS, SYF, and SFF declined (by respectively 24, 41, and 44 %). For segments with laminectomy, SS was significantly correlated with intervertebral disc degeneration and facet joint degeneration (Pfirrmann: r = 0.64; Griffith: r = 0.70; Lane: r = 0.73 and Pathria: r = 0.64), SYF was correlated with intervertebral disc geometry (r = 0.66 for length; r = 0.66 for surface and r = 0.68 for volume), BMC (r = 0.65) and frontal area (r = 0.75), and SFF was correlated with disc length (r = 0.73) and BMC (r = 0.81). For untreated segments, SS was significantly correlated with facet joint tropism (r = 0.71), SYF was correlated with pedicle geometry (r = 0.83), and SFF was correlated with BMC (r = 0.85), BMD (r = 0.75) and frontal area (r = 0.75). SS, SYF and SFF could be predicted for segments with laminectomy (r²-values respectively: 0.53, 0.81 and 0.77) and without laminectomy (r²-values respectively: 0.50, 0.83 and 0.83).

Conclusions
Significant loss of strength and SS are predicted by BMC, BMD, intervertebral disc geometry and degenerative parameters, suggesting that low BMC or BMD, small intervertebral discs and absence of osteophytes could predict the possible development of postoperative instability following lumbar laminectomy.
Introduction

Lumbar laminectomy is a commonly used treatment for symptomatic degenerative lumbar spinal stenosis. Although the impinged nerves are decompressed and neurological symptoms, such as low back pain, sciatica, claudication, motor, sensory and reflex activity, often improve following lumbar laminectomy, it can lead to symptomatic postoperative lumbar instability or even postoperative failure of the spinal motion segment. A well-known complication of lumbar laminectomy is excessive shear displacement in the intervertebral joint, leading to postoperative spondylolysis or spondylolisthesis. Symptomatic clinical instability justifies reoperation to stabilize and fuse the unstable segment. When residual strength and shear stiffness (SS) of the lumbar spine after laminectomy can be predicted, this may support patient selection for additional spinal stabilization. In other words, based on predicted residual shear properties, the surgeon may decide whether or not to combine laminectomy with (instrumented) fusion techniques.

Previously, we showed in an in vitro experiment that laminectomy resulted in a substantial decrease of SS and shear force to failure (SFF) of lumbar spinal segments. However, the biomechanical behaviour of a spinal motion segment following laminectomy will likely also depend on disc degeneration, facet joint degeneration, Modic changes, Schmorl’s nodes, intervertebral disc and pedicle geometry, and facet joint angles.

To our best knowledge, there is a lack of information in literature, demonstrating correlations between these various anatomical and clinical parameters and the biomechanical behaviour of a spinal motion segment following lumbar laminectomy. In this study, we aim to assess the relationship between various anatomical and clinical parameters, and in vitro strength and SS of a lumbar spinal segments either untreated or following facet sparing laminectomy. A total of ten spines (T12 – L5) were used. Ten segments remained untreated (five times L2 – L3 and five times L4 – L5) and ten segments were treated with laminectomy (five times L2 – L3 and five times L4 – L5).

We hypothesized that multiple independent variables, together, determine shear biomechanics of a lumbar spinal segment either intact or treated with laminectomy. Identification of these determinants may enable prediction of shear biomechanics in the future, which may support surgical decision-making.
Methods

Specimens
Thoracolumbar spines (T12 – L5) were harvested from freshly frozen (–20 °C) human cadavers (mean age 72.1 years, range 53 – 89 years). None of the deceased subjects had any history of spinal injury, spinal surgery or spinal metastatic disease. The spines were thawed before assessment and biomechanical testing. Excessive soft tissue and muscle tissue were carefully removed, keeping the anterior and posterior longitudinal ligaments as well as the facet joints intact (Figure 1).

Figure 1. Human thoracolumbar spine (T12 – L5) with a laminectomy at level L4.

Parameters
For assessment of spines, we used clinically relevant and methodologically validated parameters of lumbar spinal degeneration as recommended by the European Spine Society. Grading methods for disc degeneration with an intraclass correlation coefficient or an interobserver kappa \( \kappa > 0.60 \) \(^5, 13, 17\) were included. For facet joint degeneration, grading schemes \(^9\) with an intraclass correlation coefficient or interobserver \( \kappa > 0.40 \) were used in the present study \(^16, 26\).
Magnetic resonance imaging (MRI, Siemens Symphony 1.5 T: Syngo MR A30, software NUMARIS/4, Berlin, Germany) of lumbar spines was performed to assess intervertebral disc degeneration according to Griffith and Pfirrmann\(^5\)\(^-\)\(^7\) and facet joint degeneration according to Weishaupt\(^2\)\(^6\). Disc degeneration, (including narrowing and osteophytes, respectively Lanes 1 and 2)\(^1\)\(^3\)\(^,\)\(^2\)\(^7\), and facet joint degeneration\(^1\)\(^6\) of levels L2 – L3 and L4 – L5 were also assessed based on radiographs (Sedical Digital Vet. DX-6, Arlington Heights, IL, USA). Furthermore, MRI was used to assess the presence of Modic changes\(^1\)\(^5\) and Schmorl’s nodes\(^1\)\(^8\) and to determine intervertebral disc and pedicle geometry and facet joint angles\(^2\). Disc geometry included: disc length, width, height, surface area, and volume. Disc surface area, disc volume and pedicle diameter were calculated assuming an elliptic shape (surface = \(\frac{1}{4} \pi \times \text{length} \times \text{width}\)). For pedicle diameter, an average of left and right pedicles was taken for the top (L2 or L4) and bottom (L3 or L5) of each segment. Mean facet joint angle was calculated by averaging left and right angles per segmental level (L2 – L3 or L4 – L5) while facet joint angle differences or tropism was determined by calculating the difference between left and right facet joint angles. Segmental frontal surface area (FA), defined in \(\text{cm}^2\), bone mineral content (BMC in g) and bone mineral density (BMD in \(\text{g/cm}^2\)) of lumbar spinal sections (L2 – L3 and L4 – L5) were measured with dual X-ray absorptiometry (DXA, Hologic QDR 4500 Delphi DXA scanner, Waltham, MA, USA) in anteroposterior direction. All assessments were performed using Osirix software (Osirix, version 3.8.1., Pixmeo SARL, Geneva, Switzerland).

**Specimen preparation and biomechanical testing**

L2–L3 and L4–L5 motion segments were isolated from each spine. Subsequently, laminectomy was performed at level L2 of five randomly chosen spines, and at level L4 of the remaining five spines. Laminectomy, analogous to standard clinical practice, was performed by removing the spinous process and part of the lamina, leaving the pars interarticularis intact. During preparation, examination, and biomechanical testing, specimens were kept hydrated using 0.9 % saline-soaked gauzes. Thoracolumbar spines with bridging osteophytes, assessed on anteroposterior, lateral and oblique radiographs, were excluded from this study. After sectioning spines in L2 – L3 and L4 – L5 motion segments, the motion segments were potted in a casting-mold using low melting point (48 °C) bismuth alloy (Cerrolow – 147; 48.0 % bismuth, 25.6 % lead, 12.0 % tin, 9.6 % cadmium, and 4.0 % indium). The upper and lower vertebral bodies were fixed securely into the alloy by adding screws into the vertebral body. Screw fixation was reinforced with orthopedic bone cement (Simplex, Stryker, Kalamazoo, MI, USA). The disc was placed parallel to the flat surface of the bismuth. Discs were placed parallel based on the visual inspection. Because muscle tissue was thoroughly and carefully removed, the intervertebral disc and corresponding endplates were clearly visible. All articulating parts were kept free. The casting-mold was placed in a hydraulic materials testing machine (Instron, model 8872, Norwood, Canada)\(^1\)\(^,\)\(^2\)\(^3\)\(^,\)\(^2\)\(^4\). The caudal vertebral body was fixed on a plateau
that allowed movement in axial and transverse directions only. Transverse movements were allowed, so segments were able to find their physiological motion patterns and to correct for possible differences in embedding. Segments were loaded with a continuous axial compressive force of 1600 N, applied using a pneumatic cylinder that had been calibrated using a load cell (Hottinger Baldwin Messtechnik, Force Transducer Type C2, Darmstadt, Germany). Since compression was applied in a purely axial direction, bending moments were minimized. The level of compression simulated the force during bending, a condition in which high shear loading of the lumbar spinal segments typically occurs. Subsequently, while maintaining the axial load, anterior shear load was applied with a constant rate of 2.0 mm/min on the casting-mold containing the cranial vertebral body, until failure of the vertebral motion segment. This test set-up was similar to mechanical testing by Bisschop et al., van Solinge et al. and van Dieën et al. An anterior shear force was used since it corresponds to the loading direction in vivo. The test was stopped after hearing a crack or after a large force reduction was seen. Shear force and displacement were digitized and stored at 100 samples per second (Instron Fast Track 2, Norwood, Canada).

For each of the 20 motion segments tested, SFF was determined. SFF was defined as the point at which maximum load was recorded in the load–displacement curves for each specimen. These data were presented previously. Shear yield force (SYF) was defined as the point at which shear load caused a decrease in stiffness, i.e. a decrease in the slope of the load–displacement curve. Average SS was calculated from the load–displacement curve, between 25 and 50% of the SFF. SS was estimated by means of a least squares fit of a straight line through the data with the slope of the regression line representing stiffness. The deformation in this region was linear, with an $r^2 > 0.943$ (Table 1) between load and displacement for all motion segments. All analyses were performed using computer programs written in Matlab (Mathworks, Natick, MA, USA).
**Table 1.** Overview of specimens and biomechanical outcomes per tested segment. For shear stiffness, \( r^2 \)-values are added in brackets.

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Segment</th>
<th>Lamincetomy</th>
<th>Shear stiffness (SS) (N/mm)</th>
<th>Shear yield force (SYF) (N)</th>
<th>Shear force to failure * (SFF) (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>01 Male, 79</td>
<td>L2-L3 0</td>
<td>327 (0.998)</td>
<td>1052</td>
<td>2317</td>
<td></td>
</tr>
<tr>
<td></td>
<td>L4-L5 1</td>
<td>159 (0.999)</td>
<td>1258</td>
<td></td>
<td></td>
</tr>
<tr>
<td>02 Male, 53</td>
<td>L2-L3 0</td>
<td>213 (0.995)</td>
<td>1527</td>
<td>3284</td>
<td></td>
</tr>
<tr>
<td></td>
<td>L4-L5 1</td>
<td>247 (0.993)</td>
<td>1886</td>
<td></td>
<td></td>
</tr>
<tr>
<td>03 Male, 72</td>
<td>L2-L3 0</td>
<td>232 (0.995)</td>
<td>967</td>
<td>1678</td>
<td></td>
</tr>
<tr>
<td></td>
<td>L4-L5 1</td>
<td>307 (0.999)</td>
<td>815</td>
<td>1775</td>
<td></td>
</tr>
<tr>
<td>04 Female, 82</td>
<td>L2-L3 0</td>
<td>214 (0.988)</td>
<td>888</td>
<td>909</td>
<td></td>
</tr>
<tr>
<td></td>
<td>L4-L5 1</td>
<td>342 (0.999)</td>
<td>390</td>
<td>561</td>
<td></td>
</tr>
<tr>
<td>05 Male, 78</td>
<td>L2-L3 0</td>
<td>252 (0.998)</td>
<td>1100</td>
<td>1292</td>
<td></td>
</tr>
<tr>
<td></td>
<td>L4-L5 1</td>
<td>211 (0.997)</td>
<td>867</td>
<td>1221</td>
<td></td>
</tr>
<tr>
<td>06 Male, 79</td>
<td>L2-L3 1</td>
<td>162 (0.998)</td>
<td>431</td>
<td>994</td>
<td></td>
</tr>
<tr>
<td></td>
<td>L4-L5 0</td>
<td>378 (0.994)</td>
<td>1136</td>
<td>2408</td>
<td></td>
</tr>
<tr>
<td>07 Male, 62</td>
<td>L2-L3 1</td>
<td>200 (0.995)</td>
<td>420</td>
<td>940</td>
<td></td>
</tr>
<tr>
<td></td>
<td>L4-L5 0</td>
<td>273 (0.991)</td>
<td>1212</td>
<td>2724</td>
<td></td>
</tr>
<tr>
<td>08 Female, 64</td>
<td>L2-L3 1</td>
<td>217 (0.999)</td>
<td>304</td>
<td>660</td>
<td></td>
</tr>
<tr>
<td></td>
<td>L4-L5 0</td>
<td>236 (0.996)</td>
<td>1083</td>
<td>1553</td>
<td></td>
</tr>
<tr>
<td>09 Female, 63</td>
<td>L2-L3 1</td>
<td>64 (0.967)</td>
<td>278</td>
<td>641</td>
<td></td>
</tr>
<tr>
<td></td>
<td>L4-L5 0</td>
<td>308 (0.995)</td>
<td>1135</td>
<td>1313</td>
<td></td>
</tr>
<tr>
<td>10 Female, 89</td>
<td>L2-L3 1</td>
<td>178 (0.995)</td>
<td>709</td>
<td>721</td>
<td></td>
</tr>
<tr>
<td></td>
<td>L4-L5 0</td>
<td>309 (1.000)</td>
<td>774</td>
<td>1628</td>
<td></td>
</tr>
</tbody>
</table>

0: untreated; 1: laminectomy

* Presented previously

**Statistical methods**

Statistical analysis was performed based on two separate groups. The first group contained untreated segments (five times L2 – L3 and five times L4 – L5) while the second group consisted of segments with laminectomy (five times L2 – L3 and five times L4 – L5).

Independent variables were classified as: general variables, intervertebral disc geometry (MRI), pedicle geometry (MRI), facet joint orientation (MRI), bone characteristics (DXA), intervertebral disc degeneration classifications (MRI), intervertebral disc and facet joint degeneration (Radiographs), facet joint degeneration (MRI) and other (MRI). These classes of variables are specified in Table 2.
Table 2. Overview of correlations (P-values, two tailed < 0.05: in bold) between independent and dependent variables in untreated and treated segments. For DVs, t–values are presented while correlations based on CVs and OVs are described by Pearson’s coefficient of correlation SS shear stiffness, SYF shear yield force, SFF shear force to failure, DV dichotomized variable, CV continuous variable, OV ordinal variable.

<table>
<thead>
<tr>
<th></th>
<th>Untreated</th>
<th>Laminctomy</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>SS</td>
<td>SYF</td>
</tr>
<tr>
<td><strong>General variables</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Segment</td>
<td>DV</td>
<td>-</td>
</tr>
<tr>
<td>Sex</td>
<td>CV</td>
<td>-</td>
</tr>
<tr>
<td>Age</td>
<td>CV</td>
<td>-</td>
</tr>
<tr>
<td><strong>Intervertebral disc geometry</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Disc length</td>
<td>CV</td>
<td>MRI</td>
</tr>
<tr>
<td>Disc width</td>
<td>CV</td>
<td>MRI</td>
</tr>
<tr>
<td>Disc height</td>
<td>CV</td>
<td>MRI</td>
</tr>
<tr>
<td>Disc surface</td>
<td>CV</td>
<td>MRI</td>
</tr>
<tr>
<td>Disc volume</td>
<td>CV</td>
<td>MRI</td>
</tr>
<tr>
<td>Sections top (L2 or L4)</td>
<td>CV</td>
<td>MRI</td>
</tr>
<tr>
<td>Sections bottom (L3 or L5)</td>
<td>CV</td>
<td>MRI</td>
</tr>
<tr>
<td><strong>Facet joint orientation</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mean facet joint angle</td>
<td>CV</td>
<td>MRI</td>
</tr>
<tr>
<td>Facet joint tropism</td>
<td>CV</td>
<td>MRI</td>
</tr>
<tr>
<td><strong>Bone characteristics</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Frontal area</td>
<td>CV</td>
<td>DXA</td>
</tr>
<tr>
<td>Bone mineral content</td>
<td>CV</td>
<td>DXA</td>
</tr>
<tr>
<td>Bone mineral density</td>
<td>CV</td>
<td>DXA</td>
</tr>
<tr>
<td><strong>Intervertebral disc degeneration</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pfirrmann</td>
<td>OV</td>
<td>MRI</td>
</tr>
<tr>
<td>Griffith</td>
<td>OV</td>
<td>MRI</td>
</tr>
<tr>
<td><strong>Intervertebral disc and facet joint degeneration</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lane-1 (Narrowing)</td>
<td>OV</td>
<td>Radiographs</td>
</tr>
<tr>
<td>Lane-2 (Osteophytes)</td>
<td>OV</td>
<td>Radiographs</td>
</tr>
<tr>
<td>Wilke</td>
<td>OV</td>
<td>Radiographs</td>
</tr>
<tr>
<td>Pathria</td>
<td>OV</td>
<td>Radiographs</td>
</tr>
<tr>
<td><strong>Facet joint degeneration</strong></td>
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<td></td>
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<tr>
<td>Weishaupt</td>
<td>OV</td>
<td>MRI</td>
</tr>
<tr>
<td><strong>Other</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Schmorl’s nodes</td>
<td>DV</td>
<td>MRI</td>
</tr>
<tr>
<td>Modic changes</td>
<td>DV</td>
<td>MRI</td>
</tr>
</tbody>
</table>
First, relations between independent and dependent variables (SS, SYF and SFF) were tested for each individual variable. For dichotomized independent variables (segment, sex, Modic changes and Schmorl’s nodes, independent-sample t-tests were used while Pearson’s coefficient of correlation was determined for continuous and ordinal values. Note that it was thus assumed that ordinal variables (Pfirrmann, Griffith, Lane 1, Lane 2, Wilke, Pathria and Weishaupt) represent a linear degree of severity. When independent variables were associated with a dependent variable, here defined as independent-sample t-test: $P < 0.05$ or as a bivariate correlation with a significance level of: $P < 0.05$, they were used for the combined statistical models.

Before final analysis was performed, all independent variables were checked for correlations with each other. In case a correlation $> 0.7$ with a $P < 0.05$ was found, the independent variable with the strongest effect on the specific dependent variable was included in the model. Finally, backward linear regression techniques were used to create final statistical models per dependent variable per treatment group.

**Results**

All specimen characteristics and biomechanical properties for segments with and without laminectomy are presented in Table 1. As shown in Figure 2, lumbar laminectomy had a substantial declining effect on SS (23.7 %), SYF (41.1 %) and SFF (44.3 %).

Table 2 gives an overview of correlations between independent and dependent variables of segments with and without laminectomy. Some of the general variables, as presented in Table 1 (sex and age for untreated segments, and segment level and sex for treated segments), were related to strength parameters (SYF and/or SFF) in both groups. In the untreated segments, SFF was found to be lower for female specimens (2284 N for male versus 1351 N for female). For the treated segments, SFF was also found to be lower for female specimens (1346 N for male versus 646 N for female). SYF in treated segments proved to be level dependent (L2 – L3: 428 N versus L4 – L5: 851 N).

For segments treated with laminectomy, three out of five intervertebral disc geometry variables (i.e., length, surface and volume) were significantly related to biomechanical shear properties (SYF; all three, for SFF; length only). In contrast, biomechanics of untreated segments were unrelated to intervertebral disc geometry. The opposite was true for pedicle geometry and facet joint orientation. Pedicle sections and facet joint angle difference correlated significantly to respectively, SYF and SS in untreated segments but did not correlate with biomechanical outcomes in treated segments.
Figure 2. Effects of lumbar laminectomy on shear biomechanics, showing a substantial decrease of shear stiffness (23.7%), shear yield force (41.1%) and shear force to failure (44.3%) following laminectomy.
For both groups, bone characteristics measured with DXA, were strongly related to shear strength parameters (SYF and SFF), but not to stiffness (SS). Like intervertebral disc geometry, intervertebral disc degeneration was predictive for biomechanics (SS) of spinal segments with laminectomy. This was consistent over imaging methods and classification schemes (MRI; Pfirrmann 17 and Griffith 5 or radiographs; Wilke 27), although not significant for radiographs (correlation: 0.558, $P = 0.094$). However, in contrast to intervertebral disc geometry, these intervertebral disc degeneration scores were not related to strength (SYF and SFF). Finally, Modic changes 15 and Schmorl’s nodes 18 were not related to shear biomechanics of spinal segments with or without laminectomy. Results of the backward linear regression, using determinants of spine biomechanics, which were identified (based on a $P < 0.05$) in Table 2, are presented in Table 3. All models, describing strength parameters (SYF and SFF) consisted of two independent variables.

SYF and SFF could accurately be predicted by the final statistical model for untreated segments ($r^2$-values respectively: 0.83 and 0.83). Age and pedicle geometry remained in the model for SYF, while for SFF, the final model consisted of DXA parameters (frontal area and BMC) only. For segments treated with laminectomy, SYF and SFF could be predicted from independent variables with $r^2$-values of 0.81 (intervertebral disc volume and segment) and 0.77 (sex and intervertebral disc length), respectively. SS was less accurately predicted with only a single variable remaining in the model for both untreated segments (facet joint angle difference; $r^2 = 0.50$) and segments with laminectomy (degeneration score Lane-2 (osteophytes); $r^2 = 0.53$).
Table 3. Overview of backward linear regression models per dependent variable in untreated and in treated segments based on significant correlation coefficients found in Table 2. Each row in the table represents a regression equation. Models were based on the highest statistical power, using backward linear regression techniques.

<table>
<thead>
<tr>
<th>Untreated</th>
<th>Shear Stiffness</th>
<th>Variables:</th>
<th>Constant</th>
<th>Facet joint tropism</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>N/mm</td>
<td>Factor:</td>
<td>204</td>
<td>15</td>
</tr>
<tr>
<td></td>
<td>r²-value: 0.50</td>
<td>Significance:</td>
<td>&gt; 0.001</td>
<td>0.022</td>
</tr>
<tr>
<td>Shear yield force</td>
<td>Variables:</td>
<td>Constant</td>
<td>Age</td>
<td>Pedicle section bottom</td>
</tr>
<tr>
<td></td>
<td>N</td>
<td>Factor:</td>
<td>2102</td>
<td>-9</td>
</tr>
<tr>
<td></td>
<td>r²-value: 0.83</td>
<td>Significance:</td>
<td>&gt; 0.001</td>
<td>0.050</td>
</tr>
<tr>
<td></td>
<td>Shear force to failure</td>
<td>Variables:</td>
<td>Constant</td>
<td>Frontal area</td>
</tr>
<tr>
<td></td>
<td>N</td>
<td>Factor:</td>
<td>-2317</td>
<td>82</td>
</tr>
<tr>
<td></td>
<td>r²-value: 0.83</td>
<td>Significance:</td>
<td>0.122</td>
<td>0.093</td>
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<tr>
<td>Laminectomy</td>
<td>Shear Stiffness</td>
<td>Variables:</td>
<td>Constant</td>
<td>Lane-2 (osteoephtyes)</td>
</tr>
<tr>
<td></td>
<td>N/mm</td>
<td>Factor:</td>
<td>166</td>
<td>48</td>
</tr>
<tr>
<td></td>
<td>r²-value: 0.53</td>
<td>Significance:</td>
<td>&gt; 0.001</td>
<td>.017</td>
</tr>
<tr>
<td>Shear yield force</td>
<td>Variables:</td>
<td>Constant</td>
<td>Segment</td>
<td>Disc volume</td>
</tr>
<tr>
<td></td>
<td>N</td>
<td>Factor:</td>
<td>-47</td>
<td>363</td>
</tr>
<tr>
<td></td>
<td>r²-value: 0.81</td>
<td>Significance:</td>
<td>.773</td>
<td>.008</td>
</tr>
<tr>
<td></td>
<td>Shear force to failure</td>
<td>Variables:</td>
<td>Constant</td>
<td>Sex</td>
</tr>
<tr>
<td></td>
<td>N</td>
<td>Factor:</td>
<td>286</td>
<td>-494</td>
</tr>
<tr>
<td></td>
<td>r²-value: 0.77</td>
<td>Significance:</td>
<td>.570</td>
<td>.035</td>
</tr>
</tbody>
</table>
**Discussion**

The aim of this study was to identify parameters that predict spinal shear properties before and after laminectomy, in order to determine which of these parameters may prognosticate spinal instability following lumbar laminectomy.

For characterization of the spinal motion segments, we used commonly applied grading systems to assess disc degeneration\(^5,13,17,27^\), facet joint degeneration\(^9,16,26^\), Modic changes\(^15^\) and Schmorl's nodes\(^18^\) based on MRI and radiographs. Furthermore, we measured intervertebral disc and pedicle geometry, facet joint angles\(^2\) and bone characteristics (BMC: bone mineral content; BMD and total segmental surface area on DXA defined as frontal area: FA). These parameters all potentially affect strength and SS of the lumbar spinal motion segment before and/or after treatment with laminectomy and can be determined in clinical practice.

We showed that multiple variables are related to spinal shear properties in intact lumbar segments and lumbar segments treated with laminectomy. Statistical models with these parameters as independent variables predicted shear biomechanics, with moderate to very good accuracy with \(r^2\)-values varying from 0.50 to 0.83 (without laminectomy) and from 0.53 to 0.81 (with laminectomy). Particularly, strength parameters (SYF and SFF) in both untreated and treated segments could be predicted with good to very good accuracy. Prediction of SS was only moderately accurate.

The tests on individual variables (Table 2) showed that, for untreated segments, pedicle geometry was related to SYF and facet joint orientation to SS. In contrast, for segments with laminectomy, intervertebral disc characteristics appeared to determine shear properties. Intervertebral disc characteristics correlated to strength and disc degeneration correlated to stiffness. For both segments with and without laminectomy, DXA assessment was found to be important, although mainly for strength parameters.

SYF might be the most critical shear property, because it marks the beginning of the irreversible deformation of a spinal motion segment, signaling the appearance of the first soft tissue and or trabecular bone lesions\(^21^\). We expect that when shear loading crosses the yield point, sub-clinical damage will occur. Such damage may, at a later stage, lead to symptomatic spondylolisthesis. Unlike SYF, SFF marks, as the description suggests, complete and irreversible failure of spinal motion segments. SFF describes an acute clinically relevant situation. Therefore, SYF and SFF represent different clinical value. In untreated segments SYF depended mainly on pedicle geometry, while SFF strongly correlated with DXA parameters.
(Table 2). For treated segments, both SYF and SFF were correlated with intervertebral disc geometry and DXA parameters (Table 2) and both parameters could be predicted quite accurately (Table 3).

In this study, SS was only moderately predictable ($r^2$-values: 0.50 and 0.53). We assumed that the intervertebral disc has a large contribution to this biomechanical parameter. This assumption was corroborated by the results (Table 2). Degenerative parameters proved to be strongly correlated with SS in treated segments. In other words, laminectomy leads to a shift in load bearing, from the pars interarticularis to the intervertebral disc. Unfortunately, we could only study the morphology and degeneration of the intervertebral disc on MRI and radiographic imaging. A more specific (histological) analysis of the state of the intervertebral disc may strengthen correlations 19, but may not be clinically applicable.

For stiffness, $r^2$-values of only 0.50 and 0.53 were found. As stated earlier, stiffness was mainly determined by degenerative parameters, such as disc degeneration. The fact that these parameters are based on visual assessment and have an ordinal character possibly explains their lower predictive value, compared to directly measured continuous variables such as BMD and BMC.

In our protocol, both BMD and BMC were studied. BMD is often used as a clinical parameter. However, BMC, can also be used to express the bone mineral content since it integrates information on bone density and vertebral dimensions. BMC is defined as BMD (g/cm$^2$) multiplied by the total segmental surface area (FA) of the spinal segment (cm$^2$) and is expressed in grams. We therefore decided to include both parameters as a factor that prognosticates instability following lumbar laminectomy.

We found a substantial difference between male and female specimen considering SFF, in both treated and untreated segments. However, considering the limited number of tested specimens, we cannot draw any conclusions from these findings.

In vivo, muscle forces are very important 20. Muscle forces are the main generators of compression and shear forces. We simulated the effect of muscle forces on the spine using static 1600 N compressive force and an increasing shear force imposed by the material testing machine. The chosen preload of 1600 N was selected to allow for comparison with previous work 23, 24 and was a compromise between applying compression forces that are sufficiently large to simulate spinal loads that occur in vivo when large shear forces are present 10, 11, 12, 22, but low enough to avoid damage due to compression forces alone 3.
One limitation of this study is that small alignment errors may have been present. Our results, however, are not likely to be very sensitive to small errors in specimen alignment. Previously, it was shown that SS and SFF were not different between specimens in neutral position and specimen in 10 degrees of flexion. Therefore, we do not expect significant changes in biomechanical outcomes when malaligning segments.

Another limitation of this study was that we did not investigate the nature of failure. Van Solinge et al. investigated types of failure, occurring with shear loading. These failure mechanisms were similar to those found in clinical practice. Since our test setup was similar, we expect our segments to fail in similar fashion.

From a clinical point of view, laminectomy at a spinal segment that exhibits small intervertebral disc geometry, disc and facet joint degeneration and poor bone mineral density may need additional instrumental spinal stabilization to reduce the risk of post-operative instability. However, also pull out strength of spinal implants, proved to be dependent on bone mineral quality as measured by dual X-ray absorptiometry (DXA) and this dependency needs to be taken into account when deciding on instrumentation.

Considering further research, we recommend to assess the parameters found to be predictive in a prospective or retrospective in vivo design. In addition to shear failure, further studies should also focus on other failure mechanisms of the human lumbar spine, including axial rotation.

Finally, while $r^2$-values, as we presented, may be too low to provide the sole basis for decisions upon surgical stabilization after laminectomy. Strength parameters (SYF and SFF) correlations were predicted with reasonable accuracy ($r^2$-values between 0.77 and 0.83). As currently surgeons decide based upon personal experience, a more informed choice might benefit this decision.

In conclusion, predictive models with moderate to good accuracy were found for SYF and SFF of human lumbar spinal segments with and without laminectomy. Significant loss of SS and strength are predicted by BMC, BMD, intervertebral disc geometry and degenerative parameters. Therefore, knowledge of a patient’s BMC, BMD, intervertebral disc geometry and the possible presence of osteophytes might provide valuable information as predictors of the development of post-operative instability following lumbar laminectomy. Pedicle sections and facet geometry were not predictive for the possible development of postoperative instability following lumbar laminectomy.
References


Torsion biomechanics of the spine following lumbar laminectomy:  
a human cadaver study


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Abstract

**Purpose**
Lumbar laminectomy affects spinal stability in shear loading. However, the effects of laminectomy on torsion biomechanics are unknown. The purpose of this study was to investigate the effect of laminectomy on torsion stiffness and torsion strength of lumbar spinal segments following laminectomy and whether these biomechanical parameters are affected by disc degeneration and bone mineral density (BMD).

**Methods**
Ten human cadaveric lumbar spines were obtained (age 75.5, range 59 – 88). Disc degeneration (MRI) and BMD (DXA) were assessed. Disc degeneration was classified according to Pfirrmann and dichotomized in mild or severe. BMD was defined as high BMD (≥ median BMD) or low BMD (< median BMD). Laminectomy was performed either on L2 (5 times) or L4 (5 times). Twenty motion segments (L2 – L3 and L4 – L5) were isolated. The effects of laminectomy, disc degeneration and BMD on torsion stiffness (TS) and torsion moments to failure (TMF) were studied.

**Results**
Load–displacement curves showed a typical bi-phasic pattern with an early torsion stiffness (ETS), late torsion stiffness (LTS) and a TMF. Following laminectomy, ETS decreased 34.1 % (P < 0.001), LTS decreased 30.1 % (P = 0.027) and TMF decreased 17.6 % (P = 0.041). Disc degeneration (P < 0.001) and its interaction with laminectomy (P = 0.031) did significantly affect ETS. In the mildly degenerated group, ETS decreased 19.7 % from 7.6 Nm/degree (6.4 – 8.4 Nm/degree) to 6.1 Nm/degree (1.5 – 10.3 Nm/degree) following laminectomy. In the severely degenerated group, ETS decreased 22.3 % from 12.1 Nm/degree (4.6 – 21.9 Nm/degree) to 9.4 Nm/degree (5.6 – 14.3 Nm/degree) following laminectomy. In segments with low BMD, TMF was 40.7 % (P < 0.001) lower than segments with high BMD [34.9 Nm (range 23.7 – 51.2 Nm) versus 58.9 Nm (range 43.8 – 79.2 Nm)].

**Conclusions**
Laminectomy affects both torsion stiffness and torsion load to failure. In addition, torsional strength is strongly affected by BMD whereas disc degeneration affects torsional stiffness. Assessment of disc degeneration and BMD pre-operatively improves the understanding of the biomechanical effects of a lumbar laminectomy.
**Introduction**

Symptomatic lumbar spinal stenosis is a common degenerative disorder in the aging population. It can lead to low back pain and radiculopathy, neurogenic claudication and muscle weakness. Spinal decompression by facet joints preserving laminectomy of the affected lumbar segment is a commonly used surgical technique to alleviate symptoms. However, a decompression laminectomy obviously leads to a loss of anatomical integrity due to the removal of bony structures and the interspinous, posterior longitudinal and flavum ligaments. Despite preservation of the facet joints, lumbar laminectomy may affect spinal biomechanics, causing return of symptoms due to rotatory slips, degenerative scoliosis and post-operative fractures which are defined as post-laminectomy syndrome or failed back surgery syndrome.

The effect of lumbar laminectomy on intervertebral shear stiffness and shear force to failure is well-known. However, during daily activities such as asymmetric lifting, the lumbar spine is not only subjected to shear forces but also to torsion moments and the resulting axial rotation. Torsional injuries of the lumbar spine occur with load application accompanied by axial rotation. It is commonly held that a decreased resistance to spinal torsion is one of the most important parameters in the etiology of low back pain and disc degeneration.

In the present study, the effects of laminectomy on the torsion stiffness (TS) and torsion strength expressed as torsion moment to failure (TMF) are quantified in 20 human cadaveric lumbar spinal segments. In addition, it was also assessed whether the severity of disc degeneration and differences in bone mineral density (BMD) of the lumbar spine interact with laminectomy with respect to stiffness and failure moment.

We hypothesized that laminectomy substantially reduces TS and TMF of the human lumbar spine, and that the severity of disc degeneration and low BMD independently influence the post-operative biomechanical properties, expressed by TS and TMF, following laminectomy.

**Methods**

**Specimens and specimen preparation**

Thoracolumbar spines (T12 – L5) were harvested from freshly frozen (− 20 °C) human cadavers (mean age 75.5 years, range 59 – 88 years). None of the donors had any history of spinal injury, surgery or metastatic disease. The spines were thawed before testing. Excessive soft tissue and muscle tissue were removed, keeping the anterior and posterior...
longitudinal ligaments and facet joints intact. Lumbar spines were sectioned in an L2 – L3 and an L4 – L5 segment. To exclude systematic effects of segment level, laminectomy was performed at L2 or L4 in a balanced design. The untreated level of each thoracolumbar spine was considered as internal control. Laminectomy, analogous to standard clinical practice, was performed by removing the spinous process and attached part of the lamina and the flavum and interspinous ligaments, leaving the facet joints intact.

During preparation, assessment and biomechanical testing, specimens were kept hydrated using 0.9 % saline soaked gauzes. Furthermore, anteroposterior, lateral and oblique radiographs (Sedical Digital Vet. DX-6, Arlington Heights, IL, USA) were made to determine whether bridging osteophytes were present in segments. Thoracolumbar spines with bridging osteophytes were excluded.

Before sectioning spines in segments for testing, MRI (Siemens Symphony 1.5 Tesla: Syngo MR A30, software NUMARIS/4, Berlin, Germany) of the intact lumbar spines was performed to assess disc degeneration. Degeneration of the L2 – L3 and L4 – L5 intervertebral discs was graded according to the Pfirrmann classification of T2–weighted mid-sagittal sections. Subsequently, degeneration scores were dichotomized; grades 3 or lower were classified as ‘mildly degenerated’ while grades higher than 3 were classified as ‘severely degenerated’.

BMD (g/cm²) of each lumbar spine was determined at L1 – L4 with dual X-ray absorptiometry (DXA, Hologic QDR 4500 Delphi DXA scanner, Waltham, MA, USA) in anteroposterior direction, in accordance with common clinical practice. Low BMD was defined as lower than median, while high BMD was defined as median or higher.

After sectioning spines into L2 – L3 and L4 – L5 segments, the motion segments were potted in a casting mold using low melting point (48 °C) bismuth alloy (Cerrolow – 147; 48.0 % bismuth, 25.6 % lead, 12.0 % tin, 9.6 % cadmium and 4.0 % indium). The disc was placed parallel to the flat surface of the bismuth based on visual inspection. The upper and lower vertebral bodies were fixed securely into the alloy by adding screws into the vertebral body. Screw fixation was reinforced with orthopedic bone cement (Stryker, Simplex, Kalamazoo, MI, USA). All articulating parts were kept free.

**Biomechanical testing**

The casting mold was placed in a hydraulic materials testing machine (Instron, model 8872; Instron and IST, Norwood, Canada), to apply torsion moments. Spinal segments were tested without imposing a specific axis of axial rotation in a custom-made test setup (Figure 1).
Consequently, segments were able to find their physiological motion patterns irrespective of possible differences in embedding. During application of torsion moments, segments were loaded with a continuous purely axial compressive force of 1600 N applied using a pneumatic cylinder. Calibration of axial compression was performed using a load cell.
Baldwin Messtechnik, Force Transducer Type C2, Darmstadt, Germany). The 1600 N preload was selected to allow for comparison with load levels found in daily physiological loading and to compare with previous work, without causing compressive failure. Subsequently, torsion load was applied with a constant rate of 3.0 degrees per min by pulling on a metal wire, which was securely fixed to the part of the casting mold that contained the caudal vertebral body (Figure 1). The test was stopped after hearing a clear crack or after a large moment reduction was seen. Torsion moment and displacement were recorded and digitized at 100 Hz (Instron Fast Track 2).

For each of the 20 motion segments tested, TMF was determined. The TMF was defined as the maximum moment (in Newton meter) recorded. The torsion stiffness was calculated from the load–displacement curve. Load–displacement curves showed two distinct phases with differences in stiffness in the early and late phase of the curve. The transition phase between the early and late phase of the load–displacement curve indicated gradual yielding. Therefore, stiffness was analyzed separately for the early and late phase. Early torsion stiffness (ETS) was calculated between 20.0 and 40.0 % of the TMF, while late torsion stiffness (LTS) was calculated between 60.0 and 80.0 % of the TMF. TS was estimated by means of a least squares fit of a straight line through the torsion load–displacement data with the slope of the fitted line representing stiffness. The deformation in this region was linear between load and displacement for all motion segments. $R^2$–values were all above 0.96 except for 4 individual values (Table 1). We checked these curves visually and found that a linear fit was optimal and that the lower $r^2$–values were caused by minor irregularities in the curves rather than clear non-linearities.

**Statistical methods**

ANOVA was used to assess relationships between dependent and independent variables. Dependent variables were ETS, LTS and TMF. First, analyses were performed to determine the effect of laminectomy and degeneration on all three dependent variables, using laminectomy and dichotomized Pfirrmann scores as fixed factors and specimen as random factor. Next, we tested whether dichotomized BMD co-determined independent variables and whether these modified the effects of laminectomy, by repeating the analysis while replacing the factor dichotomized Pfirrmann score by the dichotomized BMD in the ANOVA. Note, however, that in the latter test, specimen could not be maintained as a random factor as BMD only varied between and not within segments. Consequently, this test was less sensitive for detecting effects of laminectomy, and therefore, main effects of laminectomy are not presented for this test. A significance level of 5 % was used. The statistical analyses were performed using SPSS for Mac version 16.0 (SPSS Incorporated, Chicago, IL, USA).
### Table 1. Specimens; independent and dependent variables per segment. For early and late torsion stiffness, respectively, ETS and LTS, $r^2$ values are added in brackets.

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Gender</th>
<th>Segment</th>
<th>Laminectomy</th>
<th>Disc degeneration (Pfirrmann)</th>
<th>Total bone mineral density of L1-L4 (g/cm²)</th>
<th>Early torsion stiffness (ETS) (Nm/degree)</th>
<th>Late torsion stiffness (LTS) (Nm/degree)</th>
<th>Torsion moment to failure (Nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>01</td>
<td>Male, 79</td>
<td>L2-L3</td>
<td>(0/1)</td>
<td>4</td>
<td>1.13</td>
<td>9.4 (0.998)</td>
<td>0.6 (0.854)</td>
<td>44.8</td>
</tr>
<tr>
<td></td>
<td></td>
<td>L4-L5</td>
<td></td>
<td>3</td>
<td></td>
<td>9.6 (0.442)</td>
<td>7.2 (0.986)</td>
<td>45.8</td>
</tr>
<tr>
<td>02</td>
<td>Male, 70</td>
<td>L2-L3</td>
<td>(0/1)</td>
<td>4</td>
<td>0.64</td>
<td>4.6 (0.999)</td>
<td>0.6 (0.976)</td>
<td>38.2</td>
</tr>
<tr>
<td></td>
<td></td>
<td>L4-L5</td>
<td></td>
<td>3</td>
<td></td>
<td>6.5 (0.998)</td>
<td>0.9 (0.919)</td>
<td>35.2</td>
</tr>
<tr>
<td>03</td>
<td>Male, 65</td>
<td>L2-L3</td>
<td>(0/1)</td>
<td>4</td>
<td>1.05</td>
<td>9.0 (0.989)</td>
<td>7.9 (0.998)</td>
<td>56.5</td>
</tr>
<tr>
<td></td>
<td></td>
<td>L4-L5</td>
<td></td>
<td>2</td>
<td></td>
<td>10.3 (0.997)</td>
<td>7.0 (0.998)</td>
<td>63.5</td>
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<tr>
<td>04</td>
<td>Male, 73</td>
<td>L2-L3</td>
<td>(0/1)</td>
<td>5</td>
<td>0.92</td>
<td>21.9 (0.997)</td>
<td>16.3 (1.000)</td>
<td>68.2</td>
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<td></td>
<td></td>
<td>L4-L5</td>
<td></td>
<td>5</td>
<td></td>
<td>14.3 (1.000)</td>
<td>8.5 (0.995)</td>
<td>72.5</td>
</tr>
<tr>
<td>05</td>
<td>Female, 83</td>
<td>L2-L3</td>
<td>(0/1)</td>
<td>4</td>
<td>0.70</td>
<td>18.2 (0.997)</td>
<td>9.8 (0.992)</td>
<td>46.9</td>
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<tr>
<td></td>
<td></td>
<td>L4-L5</td>
<td></td>
<td>5</td>
<td></td>
<td>8.2 (1.000)</td>
<td>3.6 (0.998)</td>
<td>29.8</td>
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<tr>
<td>06</td>
<td>Female, 83</td>
<td>L2-L3</td>
<td>(0/1)</td>
<td>3</td>
<td>0.69</td>
<td>1.8 (0.997)</td>
<td>1.9 (0.999)</td>
<td>23.7</td>
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<tr>
<td></td>
<td></td>
<td>L4-L5</td>
<td></td>
<td>3</td>
<td></td>
<td>7.9 (0.998)</td>
<td>3.4 (0.999)</td>
<td>45.0</td>
</tr>
<tr>
<td>07</td>
<td>Male, 59</td>
<td>L2-L3</td>
<td>(0/1)</td>
<td>2</td>
<td>0.81</td>
<td>3.7 (0.980)</td>
<td>1.2 (0.998)</td>
<td>43.8</td>
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<tr>
<td></td>
<td></td>
<td>L4-L5</td>
<td></td>
<td>3</td>
<td></td>
<td>8.4 (0.997)</td>
<td>1.0 (0.991)</td>
<td>56.1</td>
</tr>
<tr>
<td>08</td>
<td>Female, 84</td>
<td>L2-L3</td>
<td>(0/1)</td>
<td>3</td>
<td>0.89</td>
<td>9.3 (0.999)</td>
<td>5.6 (0.998)</td>
<td>58.3</td>
</tr>
<tr>
<td></td>
<td></td>
<td>L4-L5</td>
<td></td>
<td>4</td>
<td></td>
<td>9.6 (0.986)</td>
<td>8.3 (0.998)</td>
<td>79.2</td>
</tr>
<tr>
<td>09</td>
<td>Male, 71</td>
<td>L2-L3</td>
<td>(0/1)</td>
<td>3</td>
<td>0.55</td>
<td>1.5 (0.716)</td>
<td>0.8 (0.989)</td>
<td>24.0</td>
</tr>
<tr>
<td></td>
<td></td>
<td>L4-L5</td>
<td></td>
<td>3</td>
<td></td>
<td>6.4 (1.000)</td>
<td>2.9 (0.960)</td>
<td>27.8</td>
</tr>
<tr>
<td>10</td>
<td>Male, 88</td>
<td>L2-L3</td>
<td>(0/1)</td>
<td>4</td>
<td>0.68</td>
<td>5.6 (0.999)</td>
<td>3.3 (0.995)</td>
<td>26.9</td>
</tr>
<tr>
<td></td>
<td></td>
<td>L4-L5</td>
<td></td>
<td>4</td>
<td></td>
<td>12.1 (0.997)</td>
<td>6.4 (0.996)</td>
<td>51.2</td>
</tr>
</tbody>
</table>

0: untreated; 1: laminectomy

### Results

All specimen parameters and outcome measures are presented in Table 1. Visual inspection and MRI confirmed that facet joints were intact, and no fractures of the pars interarticularis were present in operated or intact segments before mechanical testing. Ten segments were classified as mildly degenerated and ten segments as severely degenerated. The median total BMD of all spines (L1 – L4) was 0.76 g/cm² (range 0.55 – 1.13 g/cm²). Therefore, low BMD was defined as < 0.76 g/cm² and high BMD was defined as ≥ 0.76 g/cm². Furthermore, a...
significant difference ($P < 0.001$) between mean ETS (8.9 ± 5.0 Nm/degree) and mean LTS (4.9 ± 4.1 Nm/degree) was found using a paired t-test. In some load–displacement curves, a clear yield point was seen, whereas in other, there was a gradual decline in stiffness.

**Effects of laminectomy on torsion biomechanics**

Figure 2 a presents a typical example of our data. Following laminectomy, ETS was 34.1 % ($P < 0.001$) lower than ETS in untreated segments (Figure 2b & Table 2). Mean ETS was 10.8 Nm/degree (range 4.6 – 21.9; SD 5.4 Nm/degree) in untreated segments and 7.1 Nm/degree (range 1.5 – 14.3; SD 4.1 Nm/degree) in segments with laminectomy. Following laminectomy, LTS was 30.1 % ($P = 0.027$) lower than LTS in untreated segments (Figure 2b & Table 2). Mean LTS was 5.7 Nm/degree (range 0.6 – 16.3; SD 5.0 Nm/degree) in untreated segments and 4.0 Nm/degree (range 0.8 – 8.5; SD 2.9 Nm/degree) in segments with laminectomy. Segments treated with laminectomy had a significantly lower TMF (17.6 %; $P = 0.041$) than untreated segments (Figure 2b & Table 2). Mean TMF was 51.4 Nm (range 27.8 – 79.2; SD 14.7 Nm) versus 42.4 Nm (range 23.7 – 72.5; SD 17.5 Nm) following laminectomy.

Figure 2. A. Typical example of a load–displacement curve showing the significant effects of laminectomy on early torsion stiffness (ETS; between 20–40 % of TMF), late torsion stiffness (LTS; between 60–80 % of TMF) and torsion moment to failure (TMF). The transition phase between the ETS and LTS usually reflected gradual yielding (between 40 and 60 %). In this specific example, the gradual decline in stiffness is more pronounced in the load–displacement curve of the untreated segment, than it is in the load–displacement of the treated segment. B. Schematic illustration of a load–displacement curve showing the significant effects of laminectomy on early torsion stiffness (ETS; between 20 and 40 % of TMF), late torsion stiffness (LTS; between 60 and 80 % of TMF) and torsion moment to failure (TMF). C. Schematic illustration of a load–displacement curve, showing the significant effects of disc degeneration on ETS and its significant interaction with laminectomy. D. Schematic illustration of a load–displacement curve, showing the significant effects of BMD on TMF. Significance level was presented by using asterisks (*$P < 0.05$, **$P < 0.01$ and ***$P < 0.001$).
A. Effects of laminectomy on ETS, LTS and TMF (Original data)

B. Effects of laminectomy on ETS, LTS and TMF (Schematic overview)

C. Effects of disc degeneration on ETS and its interaction with laminectomy (Schematic overview)

D. Effects of BMD on TMF (Schematic overview)
**Effects of disc degeneration on torsion biomechanics**

Segments with severe disc degeneration had a significantly higher \((P < 0.001)\) ETS than segments with mild disc degeneration (Figures 2c and 3 & Table 2). For ETS, an interaction effect \((P = 0.031)\) between disc degeneration and laminectomy was also found (Figures 2c and 3 & Table 2). Mean ETS in severely degenerated specimens was 12.1 Nm/degree (range 4.6 – 21.9; SD 6.0 Nm/degree) in untreated segments versus 9.4 Nm/degree (range 5.6 – 14.3; SD 4.5 Nm/degree) following laminectomy, equivalent to a reduction of 22.3 %. Mean ETS in the mildly degenerated group was 7.6 Nm/degree (range 6.4 – 8.4; SD 1.0 Nm/degree) in the untreated segments and 6.1 Nm/degree (range 1.5 – 10.3; SD 3.8 Nm/degree) in the treated segments, representing a reduction of 19.7 %. Note that effects of laminectomy in severely and mildly degenerated spines were smaller than in the group as a whole. This was due the fact that in the untreated group, specimens were somewhat more degenerated and degeneration did affect ETS. Considering the interaction between the effect of disc degeneration and laminectomy, it was found that the reduction of ETS following laminectomy was larger in severely degenerated discs (mean 2.7 Nm/degree or 22.3 %) than in mildly degenerated discs (mean 1.5 Nm/degree or 19.7 %). LTS was similarly affected by disc degeneration as ETS (Figure 3). However, neither the main effect of disc degeneration \((P = 0.065)\) nor its interaction with laminectomy \((P = 0.104)\) reached significance (Table 2). TMF was not affected by disc degeneration nor its interaction with laminectomy (Figure 3 & Table 2).

**Table 2.** \(P\)-values for the effects of laminectomy, as well as the effects of disc degeneration (Pfirrmann) and its interactions with laminectomy on torsion moment to failure (TMF), and early torsion stiffness (ETS; between 20 and 40 % of TMF) and late torsion stiffness (LTS; between 60 and 80 % of TMF), based on ANOVA. In addition, in the middle 2 columns the effects of BMD and its interaction with laminectomy are presented. Bold values indicate statistical significance at \(P < 0.05\).
Figure 3. The effect of laminectomy and disc degeneration on early torsion stiffness, late torsion stiffness and torsion moment to failure (mean values ± SD).
Figure 4. The effect of laminectomy and BMD on early torsion stiffness, late torsion stiffness and torsion moment to failure (mean values ± SD).
**Effects of BMD on torsion biomechanics**

Neither the main effect of low BMD nor its interaction with laminectomy did affect ETS and LTS (Figure 4 & Table 2). TMF was significantly ($P < 0.001$) higher for segments with high BMD than for segments with low BMD (Figure 2d and 3 & Table 2). For TMF, no interaction effect between BMD and laminectomy was found (Figure 4 & Table 2). In the high BMD group, mean TMF was 58.9 Nm (range 43.8 – 79.2; SD 12.1 Nm) versus a mean of 34.9 Nm (range 23.7 – 51.2; SD 10.0 Nm) in the low BMD group, representing a reduction of 40.7%.

**Discussion**

In this study, we investigated the impact of lumbar laminectomy on both torsion stiffness (TS) and TMF, and their interaction with disc degeneration and BMD.

Torsional strength of the untreated lumbar spine was studied previously $^2,9,11$. For untreated lumbar segments, we found an average TMF of 51.4 Nm, which is about twice as high as the average moment to failure reported by Adams and Hutton $^2$. These differences are most likely related to both the higher axial compression load and the free rotation center in our study. Adams and Hutton used a fixed rotation center and substantially lower axial compression loads. Furthermore, we used the ultimate force while Adams and Hutton used the yield force. Farfan et al. $^9,11$ measured TMF under compression (maximum of 573 N) in degenerated segments; their results were comparable to the present results. We used a constant compressive load level of 1600 N to allow for comparison with the load levels found in daily physiological loading $^{16}$ and to compare with previous work $^5,6$. While this force may seem high, it is not very high compared to that estimated in vivo compression. Mainly due to muscle forces, the spine is already subjected to forces of this magnitude when the trunk is inclined about 45 degrees forward. When lifting a 10 kg object from ground level, compression forces can increase up to about 5000 N $^{16}$. Failure compressive loads in human cadaveric spines are on average 3000 N $^7$. To facilitate comparison with physiologic loading conditions, spinal segments were tested without imposing a specific axis of axial rotation. Therefore, segments were able to find their physiological motion patterns irrespective of possible differences in embedding. Furthermore, we used a single loading cycle. Cyclic loading might, through visco-elastic behavior of the intervertebral disc, shift load to the posterior element $^{22}$, thereby possibly enhancing the detrimental effect of laminectomy on torsional strength. In addition, it should be noted that a limitation of our study is that we used only one specific, relatively low deformation rate (3.0 degrees/min), which might have induced some creep behaviour, effectively transferring rotational resistance from soft tissues to bony structures. However, those effects are likely small. Busscher et al. $^8$ found a 10 % increase in
axial rotation ROM after 30 min of creep, whereas in the present study TMF was reached within a few minutes in most specimens (Table 1).

We showed that laminectomy reduces TMF and TS of lumbar spinal segments by approximately 18%, 34% and 30% for ETS, LTS and TMF respectively. For shear loading, reductions in strength were larger (44.2%), while reductions in stiffness were smaller (19.9%)\(^5,6\). Reductions of these biomechanical parameters were expected since posterior elements, consisting of both soft and bony tissue, are crucial in restraining axial rotation\(^2,3,13\).

New in our study was that load–displacement curves of torsion biomechanics showed a bi-phasic pattern, with differences between stiffness in the early phase of the load–displacement curve and in the late phase of the load–displacement curve before failure (Figures 2a & 2b). Therefore, we differentiated results between ETS and LTS to separately quantify stiffness in the early and late phase of the load–displacement curve. To our best knowledge, this bi-phasic phenomenon was not described previously. We found significant differences (\(P < 0.001\)) between early and late TS of the human lumbar spine. The transition zone between ETS and LTS possibly indicates yielding. Yielding refers to a decrease in stiffness, possibly reflecting the first damage to the structure\(^20\). Since the yield-phase in the load–displacement curves did not show a well-defined transition between ETS and LTS, we could not define a yield point. We did not report axial rotation angles at failure. Figure 2a shows a typical example of data, presenting the load–displacement curve of an untreated segment. The ‘flat’ second part of the curve means that large changes in axial rotation angles can occur, without much change in torsion moment. Therefore, axial rotation angles at failure are unreliable. Concerning the failure patterns during testing, we were, unfortunately, not able to address the exact cause of failure.

It has been reported that the average torque at failure for degenerated discs is lower than for normal discs\(^11\). Our results did not corroborate this finding. However, we demonstrated a marked effect on ETS, which proved to be approximately 30% higher in severely degenerated segments in comparison to mildly degenerated treated and untreated segments. In addition, the negative effect of laminectomy on ETS was marginally larger in severely degenerated discs (mean 2.7 Nm/degree) in comparison to mildly degenerated discs (mean 1.5 Nm/degree). As expected, an interaction between disc degeneration and laminectomy was found, since laminectomy causes a shift in load-bearing from the posterior elements to anterior elements\(^14\). Notably, spinal segments treated with laminectomy and in the presence of severe intervertebral disc degeneration were still found to be stiffer than untreated mildly degenerated segments. This may reflect that severe degeneration makes the disc stiffer in torsion. Alternatively, the increased stiffness with degeneration may reflect increased
facet contact in degenerated specimens \textsuperscript{19} or direct contact of other structures, such as the endplates. Three segments were severely degenerated (Pfirrmann grade 5) and especially in these segments, results might have been affected by endplate engagement. We performed data analysis again after excluding these segments and indeed the interaction effect between ETS and degeneration disappeared after omitting these segments, while other statistical results did not change. In contrast to stiffness, we found that TMF was not affected by disc degeneration. So a stiffer intervertebral joint does not increase the failure load in torsion, which is probably due to the fact that failure occurs in the bony rather than the intervertebral disc structure.

BMD was found to have a major impact on torsion strength of spinal segments following laminectomy. The effect of low segmental BMD on TMF was even larger than the effect of laminectomy on TMF. These results were consistent with previous findings on shear strength \textsuperscript{5,6} and compressive strength \textsuperscript{7,20}. Significant effects of BMD on stiffness could not be established. Stiffness might be determined by soft tissue primarily. In addition, large standard deviations (Figure 4) might also have prevented detection of such effects. A limitation is that BMD of dissected lumbar spines is significantly lower than BMD measured in intact human cadavers \textsuperscript{21}. However, the absolute differences were small. Previously presented BMD of dissected lumbar spines was slightly higher than our BMD; however, these specimens were substantially younger \textsuperscript{15}.

Decompressive lumbar laminectomy for severe degenerative spinal stenosis usually leads to a significant postoperative relief of symptoms. Despite these good clinical results, however, some patients present themselves with recurrence of symptoms and unremitting low back pain in the long term. Radiological assessment of these patients does not show evident changes on static and dynamic radiographs, MRI or CT imaging. We hypothesize that these symptoms are a result of a post-operative change in spinal biomechanics. Yielding, i.e., passing the transition zone between ETS and LTS, may be correlated to the recurrence of symptoms. Further studies on the yielding of this typical bi-phasic stiffness of the lumbar spine are necessary. Besides defining a yield point, it could be valuable to determine what biomechanical change causes a spinal segment to yield. Factors determining the yield point and segmental stiffness were already determined for shear loading \textsuperscript{6}. Future research should focus on the prognostication of torsion biomechanics after laminectomy.

In conclusion, laminectomy affects both torsion stiffness and torsion load to failure. In addition, torsional strength is strongly affected by BMD whereas disc degeneration affects torsional stiffness. Assessment of disc degeneration and BMD pre-operatively improves the understanding of the biomechanical effects of a lumbar laminectomy.
References

Which factors prognosticate rotational spinal instability following lumbar laminectomy?

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Barend J. van Royen
Abstract

Purpose
Reduced strength and stiffness of lumbar spinal motion segments following laminectomy may lead to instability. Factors that predict shear biomechanical properties of the lumbar spine were previously published. The purpose of the present study was to predict spinal torsion biomechanical properties with and without laminectomy from a total of 21 imaging parameters.

Methods
Radiographs and MRI of ten human cadaveric lumbar spines (mean age 75.5, range 59 – 88 years) were obtained to quantify geometry and degeneration of the motion segments. Additionally, dual X-ray absorptiometry (DXA) scans were performed to measure bone mineral content and density. Facet-sparing lumbar laminectomy was performed either on L2 or L4. Spinal motion segments were dissected (L2 – L3 and L4 – L5) and tested in torsion, under 1600 N axial compression. Torsion moment to failure (TMF), early torsion stiffness (ETS, at 20 – 40 % TMF) and late torsion stiffness (LTS, at 60 – 80 % TMF) were determined and bivariate correlations with all parameters were established. For dichotomized parameters, independent–sample t-tests were used.

Results
Univariate analyses showed that a range of geometric characteristics and disc and bone quality parameters were associated with torsion biomechanical properties of lumbar segments. Multivariate models showed that ETS, LTS and TMF could be predicted for segments without laminectomy ($r^2$–values 0.69, 0.61 and 0.45, respectively) and with laminectomy ($r^2$–values 0.95, 0.87 and 0.93, respectively), with DXA–derived measures of bone quality and quantity as the main predictors.

Conclusions
Vertebral bone content and geometry, i.e. intervertebral disc width, frontal area and facet joint tropism, were found to be strong predictors of ETS, LTS and TMF following laminectomy, suggesting that these variables could predict the possible development of postoperative rotational instability following lumbar laminectomy. Proposed diagnostic parameters might aid surgical decision–making when deciding upon the use of instrumentation techniques.
Introduction

Prediction of residual strength and stiffness of a spinal segment after laminectomy is useful for a surgeon to decide whether or not to use instrumented fusion techniques. Shear biomechanical properties after laminectomy could be predicted from imaging data, which may support such decision-making. However, in daily practice the spine is also loaded in torsion. Torsion loads may cause and progress disc degeneration and may even cause failure of a segment. Previously, it has been shown in an in vitro experiment that laminectomy results in a substantial decrease of torsion stiffness and torsion moment to failure (TMF) of lumbar spinal segments. For shear loads, the biomechanical behaviour of a spinal motion segment following laminectomy has been shown to depend on disc degeneration, facet joint degeneration, Modic changes, Schmorl's nodes, intervertebral disc and pedicle geometry, and facet joint angles. This may also hold for torsion strength and stiffness following laminectomy. If true, such variables, which can be assessed based on imaging prior to surgery, may aid surgical decision-making on the need for instrumented stabilization of the spine during surgery.

Therefore, the aim of this study is to establish, as was done previously for shear biomechanics, the correlation between a broad range of spinal properties assessed by 21 imaging parameters and torsion biomechanical properties with and without laminectomy. To this end, we studied in vitro torsion stiffness and TMF of lumbar spinal segments either untreated or following facet-sparing laminectomy. Ten segments remained untreated (five times L2 – L3 and five times L4 – L5) and ten segments were treated with laminectomy (five times L2 – L3 and five times L4 – L5). We hypothesized that multiple independent variables, together, determine torsion biomechanics of a lumbar spinal segment either intact or treated with laminectomy.

Methods

Specimens

Ten lumbar spines (L1 – L5) were harvested from freshly frozen (−20 °C) human cadavers (mean age 75.5, range 59 – 88 years). The bodies were donated to the Department of Anatomy of the UMC Utrecht by last will in accordance with the Dutch legislation and were destined for medical education and research. Body handling was done according to the guidelines of the Department of Anatomy. None of the deceased subjects had any history of spinal injury, spinal surgery or spinal metastatic disease. The spines were thawed before imaging and biomechanical testing. Before imaging, also excessive soft tissues were carefully removed, keeping the anterior and posterior longitudinal ligaments as well as the facet joints intact.
Parameters

For assessment of the spines we used clinically relevant and methodologically validated parameters of lumbar spinal degeneration as recommended by the European Spine Society. Grading methods for disc degeneration with an intraclass correlation coefficient or an interobserver $K > 0.60$ were included. For facet joint degeneration, grading schemes with an intraclass correlation coefficient or interobserver $K > 0.40$ were used in the present study.

Magnetic resonance imaging (MRI, Siemens Symphony 1.5 T: Syngo MR A30, software NUMARIS/4, Berlin, Germany) of lumbar spines was performed to assess intervertebral disc degeneration according to Griffith and Pfirrmann and facet joint degeneration according to Weishaupt. Disc degeneration, narrowing (Lane–1), osteophytes (Lane–2) and facet joint degeneration of levels L2–L3 and L4–L5 were also assessed based on radiographs (Sedical Digital Vet. DX-6, Arlington Heights, IL, USA). Furthermore, MRI was used to assess the presence of Modic changes and Schmorl’s nodes and to determine intervertebral disc and pedicle geometry and facet joint angles. Disc geometry included: disc length, width, height, surface area, and volume. Disc surface area, disc volume and pedicle diameter were calculated assuming an elliptic shape. Disc geometry included: disc length, width, height, surface area, and volume. Disc surface area, disc volume and pedicle diameter were calculated assuming an elliptic shape. For pedicle diameter, an average of left and right pedicles was taken for the top (L2 or L4) and bottom (L3 or L5) of each segment. Mean facet joint angle was calculated by averaging left and right angles per segmental level (L2–L3 or L4–L5), while facet angle differences or tropism was determined by calculating the difference between left and right facet joint angles. Segmental frontal surface area (FA), defined in cm$^2$, bone mineral content (BMC, in g) and bone mineral density (BMD, in g/cm$^2$) of lumbar spinal sections (L2–L3 and L4–L5) were measured with dual X-ray absorptiometry (DXA, Hologic QDR 4500 Delphi DXA scanner, Waltham, MA, USA) in anteroposterior direction. Specimens were dissected before they were scanned in a tank filled with saline (0.9 % NaCl). All assessments were performed using Osirix software (Osirix, version 4.1.2., Pixmeo SARL, Geneva, Switzerland).

Specimen preparation and biomechanical testing

L2–L3 and L4–L5 motion segments were isolated from each spine. Subsequently, laminectomy was performed at level L2 of five randomly chosen spines, and at level L4 of the remaining five spines. Laminectomy, analogous to standard clinical practice, was performed by removing the spinous process and part of the lamina, leaving the facet joints intact. During preparation, examination, and biomechanical testing, specimens were kept hydrated using 0.9 % saline-soaked gauzes. Thoracolumbar spines with bridging osteophytes, assessed on anteroposterior, lateral and oblique radiographs, were excluded from this study. After sectioning spines in L2–L3 and L4–L5 motion segments, the motion segments were
potted in a casting-mould using low melting point \( (48 \, ^{\circ}C) \) bismuth alloy (Cerrolow – 147; 48.0 \% bismuth, 25.6 \% lead, 12 \% tin, 9.6 \% cadmium, and 4 \% indium). The upper and lower vertebral bodies were fixed securely into the alloy by adding screws into the vertebral body. Screw fixation was reinforced with orthopedic bone cement (Simplex, Stryker, Kalamazoo, MI, USA). The disc was placed parallel to the flat surface of the bismuth, based on visual inspection. Because muscle tissue was thoroughly and carefully removed, the intervertebral disc and corresponding endplates were clearly visible. All articulating parts were kept free. The casting-mould was placed in a hydraulic materials testing machine (Instron, model 8872, Norwood, Canada) \(^3\). The caudal vertebral body was fixed on a plateau that allowed movement in axial and transverse directions only. Transverse movements were allowed, so segments were able to find their physiological motion patterns and to correct for possible differences in embedding. Segments were loaded with a continuous axial compressive force of 1600 N \(^2,3,4,24,26\), a force that is compatible with lumbar spine compression forces that occur while moderately bending forward and low enough to avoid damage from compression alone \(^6\). The force was applied using a pneumatic cylinder that had been calibrated using a load cell (Hottinger Baldwin Messtechnik, Force Transducer Type C2, Darmstadt, Germany). Since compression was applied in a purely axial direction, bending moments were minimized. Subsequently, while maintaining the axial load, torsion load was applied with a constant rate of 3.0 degrees per min through a cable, which was securely fixed to the part of the casting-mould that contained the caudal vertebral body \(^3\). The test was stopped after hearing a crack or after a large moment reduction was seen. Torsion moments and displacement were digitized and stored at 100 samples per second (Instron Fast Track 2, Norwood, Canada).

For each of the 20 motion segments tested, TMF was determined. TMF was defined as the point at which maximum load was recorded in the load–displacement curves for each specimen. These data were presented previously \(^3\). Early torsion stiffness (ETS) and late torsion stiffness (LTS) were calculated from the load–displacement curve, between, respectively, 20 – 40 and 60 – 80 \% of the TMF. ETS and LTS were estimated by means of a least squares fit of a straight line through the data with the slope of the regression line representing stiffness. All analyses were performed using computer programs written in Matlab (Mathworks, Natick, MA, USA).

**Statistical methods**

Statistical analysis was performed based on two separate groups. The first group contained untreated segments (five times L2 – L3 and five times L4 – L5), while the second group consisted of segments with laminectomy (five times L2 – L3 and five times L4 – L5).
Independent variables were classified as: general variables, intervertebral disc geometry (MRI), pedicle geometry (MRI), facet joint orientation (MRI), bone characteristics (DXA), intervertebral disc degeneration classifications (MRI), intervertebral disc and facet joint degeneration (radiographs), facet joint degeneration (MRI) and other (MRI). These classes of variables are specified in Table 2.

First, relations between independent and dependent variables (ETS, LTS and TMF) were tested for each individual variable. For dichotomized independent variables (segment, sex, Modic changes and Schmorl’s nodes), an independent-samples t-test was used, while Pearson’s coefficient of correlation was determined for continuous and ordinal values. Note that it was thus assumed that ordinal variables (Pfirrmann, Griffith, Lane–1, Lane–2, Wilke, Pathria and Weishaupt) represent a linear degree of severity.

When independent variables were associated with a dependent variable, here defined as independent-samples t-test: \( P < 0.05 \) or as a bivariate correlation with a significance level of \( P < 0.05 \), they were used for the combined statistical models. Before final analysis was performed, all independent variables were checked for correlations with each other. In case a correlation \( > 0.7 \) with a \( P \)-value \( < 0.05 \) was found, the independent variable with the strongest correlation with the specific dependent variable was included in the model. Finally, backward linear regression was used to create a multivariate model per dependent variable per treatment group.

## Results

Averaged biomechanical properties for segments with and without laminectomy are presented in Table 1. Data of individual segments were presented previously.

<table>
<thead>
<tr>
<th></th>
<th>Early torsion stiffness (ETS) (Nm/degree)</th>
<th>Late torsion stiffness (LTS) (Nm/degree)</th>
<th>Torsion moment to failure (TMF) (Nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Absolute (+SD)</td>
<td>Percentage difference</td>
<td>Absolute (+SD)</td>
</tr>
<tr>
<td>Untreated</td>
<td>10.8 (5.4)</td>
<td>34.1 %</td>
<td>5.7 (5.0)</td>
</tr>
<tr>
<td>Laminectomy</td>
<td>7.1 (4.1)</td>
<td></td>
<td>4.0 (2.9)</td>
</tr>
</tbody>
</table>

Table 1. Overview of biomechanical outcomes. Early and late stiffness were estimated over 20 – 40 % and 60 – 80 % of the torsion moment to failure, respectively, and moment–deformation curves were linear within these ranges, with \( r^2 \)-values between load and displacement all above 0.96 except for four values. Bold values indicate a significance level at \( P < 0.05 \).
Table 2 gives an overview of associations between independent and dependent variables of segments with and without laminectomy. Only two of the general variables were related to biomechanical outcomes, i.e. segment for ETS in the untreated group and sex for LTS of treated segments. In the untreated segments, ETS was lower for L4 – L5 than for L2 – L3 (8.9 versus 12.6 Nm/degree). LTS in treated segments proved to be sex dependent (male 4.1 versus female 3.7 Nm/degree).

In segments with laminectomy, only ETS was related to intervertebral disc geometry variables (i.e. width and surface). In untreated segments, only TMF was related to intervertebral disc geometry (i.e. width and surface). Pedicle geometry of the top levels (i.e. L2 and L4) after laminectomy was related to ETS and TMF. Pedicle geometry of the lower levels (i.e. L3 and L5) was related to TMF in untreated segments. Facet angle difference in treated segments was correlated to both stiffness parameters (i.e. ETS and LTS).

Bone characteristics measured with DXA were strongly related with all biomechanical parameters after laminectomy. Specifically, segmental frontal area and BMC were significantly correlated to spinal biomechanics. In untreated segments, BMC was only correlated to TMF.

In contrast to DXA parameters, intervertebral disc degeneration was only predictive for biomechanics (ETS and LTS) of spinal segments without laminectomy. Furthermore, this mainly applied to MRI and radiographic classification schemes (Pfirrmann, Griffith, and Lane–1). No correlation between disc degeneration and spinal strength (TMF) was established. Finally, Modic changes and Schmorl’s nodes were only related to ETS properties of spinal segments without laminectomy.

Results of the backward linear regression, using determinants of spine biomechanics, which were identified (based on a P-value < 0.05) in Table 2, are presented in Table 3. All biomechanical parameters (ETS, LTS and TMF) could accurately be predicted by the final statistical model for segments treated with laminectomy (r²-values 0.95, 0.87 and 0.93, respectively). Intervertebral disc width, facet angle difference and BMC remained in the model for ETS, while for LTS, the final model consisted of the same parameters without the addition of intervertebral disc width. For TMF segmental frontal area and BMC defined the final model. For untreated segments, moderately predictive models for ETS, LTS and TMF were defined with r²-values of 0.69, 0.61 and 0.45, respectively. The model for ETS consisted of Pfirrmann and Modic changes as variables, while LTS was predicted by Griffith. Finally, the model for TMF consisted of pedicle geometry of the lower level.
Table 2. Overview of associations between independent and dependent variables in untreated and treated segments. ETS early torsion stiffness, LTS late torsion stiffness, TMF torsion moment to failure, DV dichotomized variable, CV continuous variable, OV ordinal variable. For DVs, t-values are presented while correlations based on CVs and OVs are described by Pearson’s coefficient of correlation. P-values are given in brackets. When P-values, two tailed > 0.05 symbols are printed bold.

<table>
<thead>
<tr>
<th></th>
<th>Untreated</th>
<th>Laminctomy</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ETS</td>
<td>LTS</td>
</tr>
<tr>
<td>Segment DV</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sex DV</td>
<td>-0.424 (0.888)</td>
<td>-0.572 (0.378)</td>
</tr>
<tr>
<td>Age CV</td>
<td>0.264 (0.461)</td>
<td>0.223 (0.535)</td>
</tr>
<tr>
<td>Disc length CV MRI</td>
<td>-0.204 (0.573)</td>
<td>0.100 (0.784)</td>
</tr>
<tr>
<td>Disc width CV MRI</td>
<td>-0.075 (0.873)</td>
<td>0.306 (0.389)</td>
</tr>
<tr>
<td>Disc height CV MRI</td>
<td>-0.127 (0.726)</td>
<td>-0.086 (0.271)</td>
</tr>
<tr>
<td>Disc surface CV MRI</td>
<td>-0.114 (0.753)</td>
<td>0.240 (0.488)</td>
</tr>
<tr>
<td>Sections top (L2 or L4) CV MRI</td>
<td>0.223 (0.536)</td>
<td>0.300 (0.400)</td>
</tr>
<tr>
<td>Sections bottom (L3 or L5) CV MRI</td>
<td>0.146 (0.687)</td>
<td>0.340 (0.336)</td>
</tr>
<tr>
<td>Mean facet angle CV MRI</td>
<td>0.254 (0.480)</td>
<td>0.334 (0.364)</td>
</tr>
<tr>
<td>Segmental frontal area CV DXA</td>
<td>-0.045 (0.903)</td>
<td>0.155 (0.710)</td>
</tr>
<tr>
<td>Bone mineral content CV DXA</td>
<td>0.319 (0.369)</td>
<td>0.376 (0.285)</td>
</tr>
<tr>
<td>Bone mineral density CV DXA</td>
<td>0.353 (0.317)</td>
<td>0.295 (0.407)</td>
</tr>
<tr>
<td>Pfirrmann OV MRI</td>
<td>0.679 (0.031)</td>
<td>0.713 (0.021)</td>
</tr>
<tr>
<td>Griffith OV MRI</td>
<td>0.650 (0.042)</td>
<td>0.781 (0.008)</td>
</tr>
<tr>
<td>Lane-1 (Narrowing) OV Radiographs</td>
<td>0.517 (0.126)</td>
<td>0.714 (0.020)</td>
</tr>
<tr>
<td>Lane-2 (Osteophytes) OV Radiographs</td>
<td>0.394 (0.260)</td>
<td>0.610 (0.061)</td>
</tr>
<tr>
<td>Weishaupt OV Radiographs</td>
<td>0.399 (0.260)</td>
<td>0.530 (0.115)</td>
</tr>
<tr>
<td>Pathria OV Radiographs</td>
<td>0.096 (0.877)</td>
<td>0.361 (0.305)</td>
</tr>
<tr>
<td>Schmorl's nodes DV MRI</td>
<td>-1.211 (0.043)</td>
<td>-1.415 (0.045)</td>
</tr>
<tr>
<td>Modic changes DV MRI</td>
<td>-1.556 (0.016)</td>
<td>-1.335 (0.319)</td>
</tr>
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</table>
Table 3. Overview of backward linear regression models per dependent variable in untreated segments, based on significant correlation coefficients found in Table 2. Each row in the table represents a regression equation.

<table>
<thead>
<tr>
<th>Untreated</th>
<th>Early torsion stiffness Nm/degree</th>
<th>Variables:</th>
<th>Constant</th>
<th>Pfirrmann</th>
<th>Modic changes</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>( r^2 )-value: 0.693</td>
<td>Factor:</td>
<td>-13.550 (6.902)</td>
<td>5.750 (1.773)</td>
<td>4.900 (2.127)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Significance:</td>
<td>0.090</td>
<td>0.014</td>
<td>0.055</td>
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<tr>
<td>Late torsion stiffness Nm/degree</td>
<td>( r^2 )-value: 0.610</td>
<td>Variables:</td>
<td>Constant</td>
<td>Griffith</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Factor:</td>
<td>-9.383 (4.394)</td>
<td>2.746 (0.775)</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Significance:</td>
<td>0.065</td>
<td>0.008</td>
<td></td>
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<tr>
<td>Torsion moment to failure Nm</td>
<td>( r^2 )-value: 0.452</td>
<td>Variables:</td>
<td>Constant</td>
<td>Pedicle sections bottom</td>
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<tr>
<td></td>
<td></td>
<td>Factor:</td>
<td>26.430 (10.380)</td>
<td>25.938 (10.100)</td>
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<tr>
<td></td>
<td></td>
<td>Significance:</td>
<td>0.034</td>
<td>0.033</td>
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<table>
<thead>
<tr>
<th>Laminectomy</th>
<th>Early torsion stiffness Nm/degree</th>
<th>Variables:</th>
<th>Constant</th>
<th>Intervertebral disc width</th>
<th>Facet angle difference</th>
<th>BMC</th>
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<tbody>
<tr>
<td></td>
<td>( r^2 )-value: 0.952</td>
<td>Factor:</td>
<td>-18.618 (4.921)</td>
<td>3.489 (0.721)</td>
<td>-0.461 (0.135)</td>
<td>0.174 (0.046)</td>
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<tr>
<td></td>
<td></td>
<td>Significance:</td>
<td>0.009</td>
<td>0.003</td>
<td>0.014</td>
<td>0.009</td>
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<tr>
<td>Late torsion stiffness Nm/degree</td>
<td>( r^2 )-value: 0.871</td>
<td>Variables:</td>
<td>Constant</td>
<td>Facet angle difference</td>
<td>BMC</td>
<td></td>
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<tr>
<td></td>
<td></td>
<td>Factor:</td>
<td>1.219 (1.817)</td>
<td>-0.493 (0.137)</td>
<td>0.166 (0.049)</td>
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<tr>
<td></td>
<td></td>
<td>Significance:</td>
<td>0.524</td>
<td>0.009</td>
<td>0.011</td>
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<tr>
<td>Torsion moment to failure Nm</td>
<td>( r^2 )-value: 0.932</td>
<td>Variables:</td>
<td>Constant</td>
<td>Segmental frontal area</td>
<td>BMC</td>
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<tr>
<td></td>
<td></td>
<td>Factor:</td>
<td>-30.540 (9.156)</td>
<td>0.898 (0.298)</td>
<td>1.300 (0.242)</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Significance:</td>
<td>0.012</td>
<td>0.019</td>
<td>0.001</td>
<td></td>
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</tbody>
</table>
Discussion

The aim of this study was to identify parameters that are correlated to spinal torsion properties, to develop a prediction model for spinal torsion instability following lumbar laminectomy.

We showed that multiple variables are related to spinal torsion properties in intact lumbar segments and in lumbar segments treated with laminectomy. Statistical models with these parameters as independent variables predicted torsion biomechanics, with moderate accuracy in untreated segments ($r^2$-values 0.45 – 0.69) and with very good accuracy ($r^2$-values 0.87 – 0.95) in treated segments.

For both treated and untreated segments, univariate associations between biomechanical behaviour and intervertebral disc geometry (i.e. disc width and disc surface) were found. It was striking that, in untreated segments, these geometrical parameters were significantly correlated to TMF, while in treated segments, a correlation with ETS was found. This possibly reflects the fact that laminectomy causes a shift in load-bearing from the posterior elements to the anterior elements, leading to a significant correlation between stiffness in the early phase and intervertebral disc geometry.

In untreated segments (Table 2), Pfirrmann and Griffith were correlated with stiffness properties (ETS and LTS), while in segments with laminectomy, no correlation was established with disc degeneration scales. While it could be argued that a difference in degeneration scores between treated and untreated segments could underlie the presence of degeneration parameters for untreated segments but not for treated segments, the data do not support this suggestion. Average degeneration scores were slightly higher for the non-treated segments, however, the standard deviation of scores was larger in treated segments.

In this study, we differentiated results between ETS and LTS in order to separately quantify stiffness in the early and late phase of the load–displacement curve. The transition zone between ETS and LTS in load–displacement curves possibly indicates yielding. This yield phase reflects a decrease in stiffness, which possibly indicates the first damage to the structure. Since the yield phase in the load–displacement curves did not show a smooth continuous curve between ETS and LTS, we could not define a specific yield point as was defined previously for shear loading. We expect that when a spinal segment reaches its LTS zone, sub-clinical damage will have occurred. Such damage may, at a later stage, lead to symptomatic spondylolisthesis and/or rotoscoliosis. Unlike LTS, TMF marks, as the description suggests, complete and irreversible failure of spinal motion segments. TMF describes an acute clinically relevant situation. Therefore, TMF and shear force to failure represent different clinical value.
For shear biomechanical prediction, DXA parameters, including BMD, BMC and segmental frontal area, were found to be of utmost importance. However, in the present study, only BMC and segmental frontal area were found to be important in the prediction of torsion stiffness parameters after laminectomy. In our protocol, both BMD and BMC were studied. BMD is often used as a clinical parameter. However, BMC can also be used to express the bone mineral content since it integrates information on bone density and vertebral dimensions. BMC is defined as BMD \( \text{g/cm}^2 \) multiplied by the total segmental surface area (FA) of the spinal segment \( \text{cm}^2 \) and is expressed in grams. Based on our results we conclude that BMC is a more useful parameter in predicting spinal biomechanical properties. Low BMC may thus be seen as an indication for using instrumentation after laminectomy. Note, however, also pull out strength of spinal implants proved to be dependent on bone mineral quality as measured by DXA and this dependency needs to be taken into account when deciding on additional instrumentation.

In vivo, muscle forces generate substantial compression forces on the spine. We simulated this by applying a static 1600 N compressive force. Mainly due to muscle forces, the spine is already subjected to forces of this magnitude when the trunk is inclined about 45 degrees forward. When lifting a 10 kg object from ground level, compression forces can increase up to about 5000 N. Failure compressive loads in human cadaveric spines are on average 3000 N. The compression load of 1600 N was also selected to allow for comparison with previous work and was sufficiently large to simulate physiological loading, but low enough to avoid damage due to compression forces alone.

Finally, while \( r^2 \)-values, as we presented for untreated segments, may be too low (\( r^2 \)-values 0.45 – 0.69) to provide clinically valuable data for the untreated segment, biomechanical parameters in treated segments were predicted with high accuracy (\( r^2 \)-values 0.87 – 0.95). At present, surgeons decide based upon personal experience whether to apply instrumented stabilization after laminectomy. Our prediction models for laminectomy segments suggest that clinical decision-making may benefit from taking variables into account that are readily available from DXA measurements, i.e. segment frontal area and BMC. Considering further research, we recommend to assess the parameters, found to be predictive here, in a prospective or retrospective in vivo design, to define true clinical value. Furthermore, we were not able to obtain younger specimen with high DXA scores and completely healthy discs for our study. Therefore we must acknowledge that our data applies to the elderly population and is may not hold for younger subjects.
In conclusion, predictive models, mainly based on DXA variables, were found to accurately predict rotational biomechanical behaviour of the human lumbar spinal segments treated with laminectomy. In contrast, biomechanical behaviour of untreated segments was only moderately predicted. Proposed diagnostic parameters might aid surgical decision-making when deciding upon the use of instrumentation techniques.
References

Effects of repetitive movement on range of motion and stiffness around the neutral orientation of the human lumbar spine


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Abstract

Purpose
In loading experiments on the lumbar spine, typically three consecutive loading cycles are applied of which the third cycle is used for analysis. The aim of this study was to investigate whether the use of ten instead of three loading cycles reduces effects of viscoelastic behavior in the assessment of range of motion (ROM) and stiffness around the neutral orientation of the human lumbar spine.

Methods
To this end, twelve cadaveric human lumbar spines (L1 – L5) were obtained (mean age: 76.9 years, range 59 – 90 years). Before testing, the spines were subjected to a compressive load of 250 N for 1 hour. To each spine, ten consecutive loading cycles were applied (–4 Nm to +4 Nm) in flexion and extension (FE), lateral bending (LB) and axial rotation (AR). The ROM and stiffness within the neutral zone were calculated per motion segment (L2 – L3, L3 – L4 or L4 – L5) from load–displacement data.

Results
It was found that the ROM increased significantly (all $P < 0.001$) in all directions after three (FE: 0.07 degree / 1.0 %, LB: 0.08 degree / 1.5 %, and AR: 0.04 degree / 1.5 %) and after ten loading cycles (FE: 0.20 degree / 2.9 %, LB: 0.16 degree / 3.3 %, and AR: 0.09 degree / 3.3 %). Stiffness was not significantly affected, but varied considerably over cycles.

Conclusions
Although effects were small, assessment of the tenth cycle instead of the third cycle reduces viscoelastic effects in repeated measurements of ROM, because the spine is closer to a steady state condition, while averaging over loading cycles would improve the assessment of stiffness estimates.
Introduction

Spinal segments are subjected to repeated daily movements and loading by gravity and muscle forces\(^\text{15}\). Previously it has been shown that mechanical properties of spinal segments are time–dependent, due to its visco– and poro–elastic properties\(^\text{8,9,12,16}\). Changes in spinal biomechanics are therefore likely to occur when repeated loading cycles are applied. These may affect biological behavior during the day and may act as a confounder in experimental protocols. To our knowledge, little is known about the effects of repetition in biomechanical testing of the human lumbar spine.

Previously, it has been recommended to perform a set of three consecutive loading cycles and to analyze the third cycle in spinal testing\(^\text{6,14}\) and most studies have followed this guideline.

The aim of this study was to investigate whether the use of ten instead of three loading cycles would improve preconditioning of the spine and reduce effects of viscoelastic behavior in the assessment of the range of motion (ROM) and stiffness around the neutral orientation of the human lumbar spine.

Methods

Twelve lumbar spines (L1 – L5) were harvested from freshly frozen (–20 °C) human cadavers (mean age 76.9 years, range 59 – 90 years). Spines with bridging osteophytes seen on radiographs were excluded. The spines were thawed before testing. Excessive soft tissue and muscle tissue was carefully removed, keeping the anterior and posterior longitudinal ligaments, flavum ligaments, interspinous ligaments and supraspinous ligaments as well as the facet capsular ligaments intact. Specimens were kept hydrated using 0.9 % saline-soaked gauzes. After radiographic assessment, spinal segments (L1 and L5) were potted in a casting–mold and partially buried in a low melting point (48 °C) bismuth alloy. All articulating parts were kept free. Preparation of specimens was extensively described previously\(^\text{2,3}\).

The test setup was described previously\(^\text{4,5}\). Lumbar spines were placed horizontally in a custom made four–points bending device in which pure moments in flexion and extension (FE) lateral bending (LB) and axial rotation (AR) can be applied, using a hydraulic materials testing machine (Instron, model 8872; Instron and IST, Norwood, Canada). Markers containing three LEDs were screwed to the anterior surface of the vertebral bodies of L2, L3 and L4 and to the L5 casting–mold.
Before testing, a compressive preload of 250 N was applied for 1 hour. This axial preload was selected to allow comparison with previous work \cite{4, 5} and to minimize the risk of buckling of the whole lumbar spine. Mechanical testing started immediately after the preloading period. During testing, no compressive load was applied, again to prevent buckling. Loads were increased to 4 Nm \cite{4, 5, 7, 14} at an angular velocity of 0.5 degrees/sec \cite{13, 14}. At 4 Nm, loading was decreased, again at 0.5 degrees/sec, to reach –4 Nm. Each movement direction was tested for ten consecutive cycles. Force and displacement of the Instron were recorded and digitized at 100 Hz (Instron Fast Track 2). All tests were performed at room temperature. The first six segments were tested in the order: FE–LB–AR while the second six segments were tested in the order: AR–FE–LB, to correct for order effects. Time between different loading directions, needed for converting the test setup, was approximately 5 minutes.

Motions of the LEDs on L2, L3, L4 and the L5 casting–mold were recorded at 100 samples/s by a motion analysis system (Optotak 3020, Northern Digital Inc., Waterloo, ON). Using measured forces and LED displacements, a computer program (Mathworks, Natick, MA, USA) calculated the load–displacement curves in the loaded direction for L3 relative to L2, for L4 relative to L3 and for L5 relative to L4, to quantify the behavior of segments L2 – L3, L3 – L4 and L4 – L5.

For each direction (FE, LB and AR) the ROM (degrees) and stiffness (Nm/degree) per motion segment (L2 – L3, L3 – L4 and L4 – L5) were calculated from load–displacement data using Matlab (Mathworks, Natick, MA, USA). The ROM was calculated between an applied load of –4 Nm and +4 Nm. Stiffness was estimated as the slope of a least squares straight line fit through load–displacement data within the neutral zone \cite{11}. When the $r^2$–value of this fit was below 0.95, stiffness was calculated between –1.0 Nm and +1.0 Nm.

To test the effect of repetitive movements, a repeated measures analysis of variance (ANOVA) was performed with load cycle (using all ten cycles) as a within-subject factor and spinal level as a between-subject factor. In case of significant effects of cycle, we performed planned comparisons (paired $\tau$–tests) between all subsequent cycles and, additionally, between cycles one and three, cycles three and ten and cycles one and ten. Load sequence effects were tested on the average of ten cycles, using unpaired $\tau$–tests. The statistical analyses were performed using SPSS for Mac version 20.0 (SPSS Incorporated, Chicago, IL, USA).
Results

ROM and stiffness in all loading directions of L2 – L3; specimen 08, L2 – L3; specimen 09 and L4 – L5; specimen 12, were excluded from analysis due to severely irregular load–displacement curves. Regarding stiffness, linearity of the fit in 16 of the remaining 198 determinations did not reach an $r^2$–value < 0.95 and for these fits were made on data obtained between −1 Nm and +1 Nm Table 1.

ROM in LB was affected by load sequence ($P = 0.03$), while FE ($P = 0.27$) and AR ($P = 0.58$) were not. No significant effects of load sequence on stiffness (FE: $P = 0.17$; LB: $P = 0.26$ and AR: $P = 0.25$) were found.

ROM was significantly affected by cycle in all directions (FE: $P < 0.001$; LB: $P < 0.001$ and FE: $P < 0.001$; Figure 1 & Table 2). No significant effects of segment level or interactions with segment level were found (Table 2). Figure 1 and Table 3 show a significant increase in ROM between the first and third (FE: 0.07 degree / 1.0 %, LB: 0.08 degree / 1.5 %, and AR: 0.04 degree / 1.5 %), between the third and tenth (FE: 0.13 degree / 1.9 %, LB: 0.08 degree / 1.8 %, and AR: 0.05 degree / 1.8 %) and between the first and tenth load cycle (FE: 0.20 degree / 2.9 %, LB: 0.16 degree / 3.3 %, and AR: 0.09 degree / 3.3 %).

Stiffness was not affected by repetitive movement (Figure 2 & Table 2) and no significant effects of segment level or interactions with segment level were found (Table 2).
Table 6.

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Range of Motion (degrees)</th>
<th>Stiffness (Nm/degree)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>FE</td>
<td>LB</td>
</tr>
<tr>
<td>01</td>
<td>9.40</td>
<td>9.51</td>
</tr>
<tr>
<td>02</td>
<td>7.89</td>
<td>7.98</td>
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<tr>
<td>03</td>
<td>7.41</td>
<td>7.50</td>
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<td>6.47</td>
<td>6.59</td>
</tr>
<tr>
<td>05</td>
<td>8.08</td>
<td>8.24</td>
</tr>
<tr>
<td>06</td>
<td>8.40</td>
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<td>07</td>
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<td>3.55</td>
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<tr>
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<td>3.00</td>
<td>3.00</td>
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<td>4.72</td>
</tr>
<tr>
<td>12</td>
<td>4.00</td>
<td>4.13</td>
</tr>
</tbody>
</table>

**Range of Motion and Stiffness, averaged over spinal levels and per specimen of the first, third and tenth cycle. An average over L2 – L3, L3 – L4 and L4 – L5 was used (L2 – L3 excluded for analysis; L4 – L5 excluded for analysis).**
Figure 1. Effects of repetitive movement on range of motion per spinal segment and also as an average of all spinal segments (mean ± SD). In addition, normalized values per cycle, averaging 12 specimens, are presented below. Significant differences between cycles are now shown (* P < 0.05; ** P < 0.01 and *** P < 0.001) (Table 3). Note that P-values were based on paired t-tests including absolute values instead of normalized values.

Table 2. ANOVA outcomes for the range of motion and stiffness, cycles, level and the interaction between cycle and level are shown.

<table>
<thead>
<tr>
<th></th>
<th>FE</th>
<th>LB</th>
<th>AR</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>ROM (degrees)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cycle</td>
<td>P &lt; 0.001</td>
<td>P &lt; 0.001</td>
<td>P &lt; 0.001</td>
</tr>
<tr>
<td>Level</td>
<td>P = 0.419</td>
<td>P = 0.719</td>
<td>P = 0.592</td>
</tr>
<tr>
<td>Cycle x Level</td>
<td>P = 0.870</td>
<td>P = 0.265</td>
<td>P = 0.490</td>
</tr>
<tr>
<td><strong>Stiffness (Nm/degree)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cycle</td>
<td>P = 0.632</td>
<td>P = 0.545</td>
<td>P = 0.273</td>
</tr>
<tr>
<td>Level</td>
<td>P = 0.425</td>
<td>P = 0.528</td>
<td>P = 0.405</td>
</tr>
<tr>
<td>Cycle x Level</td>
<td>P = 0.715</td>
<td>P = 0.396</td>
<td>P = 0.691</td>
</tr>
</tbody>
</table>
Figure 2. Effects of repetitive movement on stiffness per spinal segment and also as an average of all spinal segments (mean + SD). In addition, normalized values per cycle, averaging 12 specimens, are presented below. No significant differences between cycles were found.
Table 3. Follow-up planned comparison showing statistical outcomes of paired t-tests between cycles. Percentage differences between the first and third; the third and tenth and the first and tenth cycle were presented with corresponding \(P\)-values. The use of this technique is justified since there were no significant effects for level or the interaction between level and cycle (Table 2).

<table>
<thead>
<tr>
<th>ROM</th>
<th>FE</th>
<th>LB</th>
<th>AR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cycle</td>
<td>(\Delta)</td>
<td>(P)-values</td>
<td>(\Delta)</td>
</tr>
<tr>
<td>1 vs. 3</td>
<td>+1.0 %</td>
<td>(P = 0.005)</td>
<td>+1.5 %</td>
</tr>
<tr>
<td>3 vs. 10</td>
<td>+1.9 %</td>
<td>(P = 0.001)</td>
<td>+1.8 %</td>
</tr>
<tr>
<td>1 vs. 10</td>
<td>+2.9 %</td>
<td>(P &lt; 0.001)</td>
<td>+3.3 %</td>
</tr>
</tbody>
</table>

**Discussion**

We studied the effects of repetitive movement in FE, LB and AR on ROM and stiffness in twelve human cadaveric lumbar spines.

Repetitive movement increased ROM significantly after three and ten consecutive loading cycles, while stiffness was unaffected. Therefore, it seems that only ROM, in all three motion directions, is influenced by visco- and/or poroelastic properties of the human lumbar spine. Previously, Wilke et al. 13 showed that ambient conditions are of greater influence on ROM than cycle count is. On the other hand, Panjabi et al. 10 found no differences in ROM after testing his spines for a limited number of cycles in the morning and afternoon over 13 consecutive days with refrigerated storage in between tests. Ambient exposure time in our study was much shorter than in both previous studies.

It has been recommended to use pure moments of 7.5 Nm in testing lumbar spines and about half of that when testing osteoporotic spines 14. Since we tested spines of elderly donors, which could be osteoporotic, we decided to use a 4 Nm load level. Comparable load levels were previously used 4, 5, 7.

One limitation of this study is that we used a preload of 250 N for 1 hour, which is relatively low in perspective of physiological loading 1. Preloading with 250 N was chosen, since buckling might occur when loading a complete lumbar spine with a higher load. Another limitation is that we did not apply axial loading during testing, again to prevent buckling. While so-called follower loads would allow for axial loading during bending without buckling, such loads inevitably cause additional moments of unknown magnitude, which would interfere with the purpose of the present study. Furthermore, we did not apply complex 3D motions to the spines, whereas during daily life, the spine is often subjected to a combination of different loading directions. We tested ten repetitive cycles only, but visual inspection (Figure
1) showed that the effect of repetitive movement for ROM reached a stable phase within ten cycles, while for stiffness, only apparently random variation was observed (Figure 2). Finally, we only used an angular velocity of 0.5 degrees per second. This velocity is commonly used and fairly low, for this reason we believe that a moderate change of this velocity would not affect our results.\textsuperscript{13, 14}

ROM showed a small but significant increase between the first and third cycle (0.04 – 0.08 degrees), but also between the third and tenth cycle (0.05 – 0.13 degrees). Wilke et al.\textsuperscript{14} stated that at least two precycles before testing are necessary to minimize viscoelastic effects, because load displacement behavior of the first two cycles could clearly be distinguished, whereas the difference between the second and third cycles was considerably reduced and the third cycle was in many cases nearly identical to all subsequent cycles. We have shown that viscoelastic effects do not stop with three cycles. At ten cycles, ROM curves are closer to an asymptote than at three cycles (Figure 1). Therefore, in experiments that rely on repeated ROM measurements, preconditioning with ten cycles instead of three would be indicated to reduce confounding effects of viscoelasticity. However, the magnitude of the viscoelastic effects, both between one and three cycles and between three and ten cycles, is small and it may depend on the specific goal of the experiments whether or not these effects are important.

Figure 2 shows that spinal stiffness varies considerably. As for ROM, it has also been recommended to analyze the third of three subsequent loading cycles.\textsuperscript{6, 14} Since we did not find clear changes in stiffness over repeated load cycles, our results imply that averaging stiffness values over three cycles, or better over ten cycles, would improve the precision of stiffness estimates.

In conclusion, using ten instead of three cycles reduces viscoelastic effects in repeated measurements, because the spine is closer to a steady state ROM condition. Averaging stiffness values over three, or preferably ten, loading cycles improves the assessment of spinal stiffness.
References

Single level lumbar laminectomy alters the biomechanical behavior of the treated segment without affecting adjacent segments

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Abstract

Purpose
Degenerative lumbar spinal stenosis causes neurological symptoms due to neural compression. Lumbar laminectomy is a commonly used treatment for symptomatic degenerative spinal stenosis. However, it is unknown if and to what extent single level laminectomy affects the range of motion and stiffness of treated and adjacent segments. An increase in range of motion and a decrease in stiffness are possible predictors of post-operative spondylolisthesis or spinal failure.

Methods
Twelve cadaveric human lumbar spines were obtained. After preloading, spines were tested in flexion-extension, lateral bending, and axial rotation. Subsequently, single level lumbar laminectomy analogous to clinical practice was performed at level lumbar 2 or 4. Thereafter, load–deformation tests were repeated. The range of motion and stiffness of treated and adjacent segments were calculated before and after laminectomy. Untreated segments were used as control group. Effects of laminectomy on stiffness and range of motion were tested, separately for treated, adjacent and control segments, using repeated measures analysis of variance.

Results
Range of motion at the level of laminectomy increased significantly for flexion and extension (7.3 %), lateral bending (7.5 %), and axial rotation (12.2 %). Range of motion of adjacent segments was only significantly affected in lateral bending (−7.7 %). Stiffness was not affected by laminectomy.

Conclusions
The increase in range of motion of 7 – 12 % does not seem to indicate the use of additional instrumentation to stabilize the lumbar spine. If instrumentation is still considered in a patient, its primary focus should be on re-stabilizing only the treated segment level.
Introduction

In neurosurgical and orthopedic practice, elderly patients often present with symptomatic degenerative lumbar spinal stenosis. A commonly used surgical decompression procedure for this type of spinal stenosis is a single level facet-sparing laminectomy. Although the impinged nerves are decompressed and neurological symptoms, such as sciatica, claudication, and motor-, sensory- and reflex activities, improve following lumbar laminectomy, the anatomically destructive character of this technique obviously affects spinal biomechanics. In fact, laminectomy can lead to symptomatic postoperative lumbar clinical instability i.e. spondylolisthesis or even postoperative failure of the spinal motion segment. Spondylolisthesis is the forward motion of a spinal segment with respect to its underlying segment. Spondylolysis or post-operative failure includes the fracturing of the posterior arch, facet joint and/or vertebral body. Both spondylolisthesis and spondylolysis generally occur post-operatively. Symptomatic clinical instability justifies re-operation to stabilize and fuse the unstable segment. The incidence of iatrogenic spondylolisthesis after facet sparing laminectomy has been reported to range from 8 to 31 %.

Previously it has been shown in vitro that facet-sparing laminectomy reduces the threshold at which shear forces and torsion moments cause lumbar spinal failure. It seems plausible that this type of decompressive surgery could also affect spinal biomechanics under submaximal loads. Reduced stiffness and increased range of motion under sub maximal loading might make the segment more vulnerable and could lead to large tissue strains, impingements, or even tissue failure such as in iatrogenic spondylolisthesis and spondylolysis. Moreover, facet-sparing laminectomy could affect, due to its effects on the anatomical integrity of the lumbar spine, not only the treated segments but also adjacent segment levels. At present, little is known about the effects of single level facet-sparing lumbar laminectomy on the flexibility of the whole lumbar spine, i.e., on the range of motion (ROM) and stiffness in flexion–extension (FE), lateral bending (LB) and axial rotation (AR). Such effects are likely smaller than biomechanical changes caused by more extensive or multilevel decompressive surgery.

In the present study, we quantified the effects of facet sparing single level laminectomy on the Rom and stiffness of lumbar (L) levels L2 – L3, L3 – L4 and L4 – L5 in FE, LB and AR under sub-maximal loading, using twelve fresh frozen human cadaveric lumbar spines (L1 – L5). We hypothesized that laminectomy causes an increase in the ROM and a reduction of stiffness of the treated segment while affecting adjacent segments of the lumbar spine to a lesser extent or not at all. An increase in ROM and a decrease of stiffness could, through a mechanism of cumulative damage, ultimately result in post-operative tissue failure, both in bone and soft tissue.
Methods

Specimens and specimen preparation

Twelve lumbar spines (L1 – L5) were harvested from freshly frozen (−20 °C) human cadavers (mean age 76.9 years, range 59–90 years). The bodies were donated to the Department of Anatomy of the UMC Utrecht by last will in accordance with the Dutch legislation and were destined for medical education and research. Body handling was done according to the guidelines of the Department of Anatomy. None of the deceased subjects had any history of spinal injury, spinal surgery or spinal metastatic disease. The spines were thawed before testing. Excessive soft tissue and muscle tissue was carefully removed, keeping the anterior and posterior longitudinal ligaments as well as the facet joints intact. During preparation, assessment and biomechanical testing, specimens were kept hydrated using 0.9% saline-soaked gauzes. Anteroposterior, lateral and oblique radiographs (Sedical© Digital Vet. DX-6, Arlington Heights, IL, USA) were made to determine whether bridging osteophytes were present in segments. Lumbar spines were considered eligible for this study in case no bridging osteophytes were seen. Magnetic resonance imaging (MRI, Siemens© Symphony 1.5 T: Syngo MR A30, software NUMARIS/4, Berlin, Germany), lateral and oblique radiographs and visual inspection also confirmed that facet joints were intact and no fractures of the pars interarticularis were present in segments before mechanical testing.

After imaging assessment, the top and bottom vertebrae of the lumbar spine (L1 and L5) were potted in a casting–mold and partially buried in a low melting point (48 °C) bismuth alloy (Cerrolow–147; 48.0 % bismuth, 25.6 % lead, 12.0 % tin, 9.6 % cadmium and 4.0 % indium) (Figure 1a). The L1 and L5 vertebral bodies were firmly fixed into the alloy by adding screws into the vertebral body prior to submerging in the alloy. Discs of the top and bottom vertebra were placed parallel based on a visual inspection. Because the muscle tissue was thoroughly and carefully removed, the intervertebral disc and corresponding endplates were clearly visible. All articulating parts were kept free. Markers containing three LED’s were rigidly fixed with screws to the anterior surface of the vertebral bodies of L2, L3 and L4 and to the casting-mold in which L5 was mounted (Figure 2).
Figure 1a. Untreated spinal segment. Both L1 (shown left) and L5 (shown right) are potted in a casting-mold using a Bismuth-alloy.

Figure 1b. The same spinal segment as shown in panel 1a after performing a single level laminectomy on level L4. The spinous process, supraspinous and interspinous ligaments are removed while the facet-joint remained intact.

Figure 2. Test setup for the application of continuous moments to the multilevel spinal segments. The specimen was rotated 90° for lateral bending. For axial rotation, the left side with the cup was rotated with a steel cable driven by the same mechanical testing system (not shown in this figure). The markers were rigidly fixed to the vertebrae. An Optotrak camera registered the movement of the LED’s. The segment as shown was untreated. Markers on the casting-mold, containing L1 and L5 are not shown in this figure.
Biomechanical testing

The test setup was similar to previous studies \(^4,6,7,26\). Segments were placed horizontally in a custom made four-point bending device in which FE, LB and AR were applied using a hydraulic materials testing machine (Instron©, model 8872; Instron and IST, Norwood, Canada). This setup guarantees that forces generate a moment that is equal at all levels of the lumbar spine. During and in between testing procedures, spines were kept hydrated by covering them in saline-soaked gauzes.

Before testing, a compressive preload of 250 N was applied for one hour. A pure axial compressive force was applied using a pneumatic cylinder. Calibration of axial compression was performed using a load cell (Hottinger Baldwin Messtechnik©, Force Transducer Type C2, Darmstadt, Germany). The chosen amount of axial preload, which is somewhat below the load of bodyweight and muscle forces in upright posture, was selected to allow for comparison with previous work \(^4,6,7,26\) and to minimize the risk of buckling of the whole lumbar spine during preloading. During testing, no compressive load was applied, again, in order to prevent buckling of the multi-segmented spine \(^22\). Loads were applied up to a moment of 4 Nm at a constant angular velocity of 0.5 degree/sec \(^29\). When a moment of +4 Nm was measured, the Instron reversed its loading direction until −4 Nm was reached. Each movement direction was tested for ten consecutive cycles \(^4\). Force and displacement of the Instron were recorded and digitized at 100 Hz (Instron© Fast Track 2, Norwood, Canada). All tests were performed at room temperature.

After the first set of measurements (FE, LB and AR), which took approximately 15 – 20 min, laminectomy was performed at level L2 of six randomly chosen lumbar spines and at level L4 of the remaining six lumbar spines. Laminectomy was performed by removing the spinous process and part of the lamina while leaving the facet joints intact (Figure 1b). By using this technique, the integrity of supraspinous, interspinous and flavum ligaments is also lost. This technique is analogous to the standard clinical practice. Again, a compressive preload of 250 N was applied for 1 h. Thereafter spinal segments were tested in FE, LB and AR for another set of ten consecutive cycles. To correct for a possible systematic effect of test sequence the order of testing was varied over spines: the first six segments (three times laminectomy at level L2 and three times laminectomy at level L4) were tested in the order FE–LB–AR–laminectomy–AR–LB–FE while the remaining six segments (three times laminectomy at level L2 and three times laminectomy at level L4) were tested in the order AR–FE–LB–laminectomy–LB–FE–AR.

During testing, motions of the LEDs on L2, L3, and L4 and the casting mold containing L5 were recorded by an optoelectronic three dimensional movement registration system with one array of three cameras (Optotrak© 3020, Northern Digital Inc., Waterloo, ON, Canada).
Labview software was used for data acquisition. The sampling rate was 100 samples per second. The three-dimensional precision of this system at a distance of 2 m is about 0.01 mm. Before testing, the axes of the Optotrak system were aligned with the anatomic axes of the lumbar spine. Using the applied moment and Optotrak LED displacements a Matlab (Mathworks Inc., Natick, MA, USA) computer program calculated the load–displacement curves in the loaded direction for L3 relative to L2, for L4 relative to L3 and for L5 relative to L4. Subsequently, the biomechanical behavior (i.e. ROM and stiffness) of three segments was analyzed (L2 – L3, L3 – L4 and L4 – L5).

**Data analysis**

Laminectomy was performed on a total of twelve segments (six times at level L2 and six times at level L4). The opposite untreated segments (six times segment L4 – L5 and six times segment L2 – L3, respectively) were used as control group. For each individual test (FE, LB and AR) the ROM (degrees) and stiffness (Nm/degree) before and after laminectomy per motion segment (L2 – L3, L3 – L4 and L4 – L5) were calculated from the load–displacement data using Matlab (Mathworks©, Natick, MA, USA). The ROM was calculated from a double sigmoid curve fitted through the load–displacement data between an applied load of −4 Nm and +4 Nm 25 (Figure 3). The tenth cycle was used for analysis 4. Stiffness was estimated by means of a least squares fit of a straight line through a section of the fitted curve between −1.0 Nm and +1.0 Nm with the slope of the line representing stiffness. For stiffness an average of ten cycles was used 4. For both stiffness and ROM, upward curves (i.e. from −4 Nm to +4 Nm) were used.

**Figure 3.** Typical example of a load–displacement curve showing displacement (in degrees) on the x-axis and load (in Nm) on the y-axis. The blue line represents original data, while the pink line represents the fitted curve. At −1 Nm and +1 Nm two vertical lines are drawn to indicate that spinal stiffness is measured within this load-region.

**Statistical methods**

Effects of laminectomy on stiffness and range of motion were tested, separately for treated, adjacent and control segments, using repeated measures analysis of variance. Pre-post treatment was used as within subjects factor. For treated and control segments, segment level (L2 – L3 or L4 – L5) was used as between subjects factor in the ANOVA. Adjacent segments
were always levels L3 – L4, but laminectomy had been either applied proximally (L2 – L3) or distally (L4 – L5). Therefore, for adjacent segments, treatment level rather than segment level was used as between subjects factor. A significance level of 5% was used. In case of significant interactions, Bonferroni-corrected paired t-tests were applied. The statistical analyses were performed using SPSS for Mac version 20.0 (SPSS Inc.©, Chicago, IL, USA).

Results

An overview of specimen characteristics is presented in Table 1. All statistical results are presented in Tables 2 and 3. ROM and stiffness in all motion directions of specimen 08 (L2 – L3), specimen 09 (L2 - L3) and specimen 12 (L4 – L5) were excluded from the analysis, due to severely irregular load displacement curves. For ROM, 4 out of the remaining 198 analyzed cycles did not reach a fit of the double sigmoid curve of $r^2 > 0.95$ (range: 0.914 – 0.939). These cycles were individually assessed for quality; no cycles were excluded. For stiffness, 67 of the remaining 1980 measurements did not reach a linear fit of $r^2 > 0.95$ between $-1.0$ and $+1.0$ Nm, and were, after individual assessment, all excluded from the analysis due to severe irregularities in the data between $-1.0$ and $+1.0$ Nm.

Table 1. Overview of specimens.

<table>
<thead>
<tr>
<th>Donor</th>
<th>Gender</th>
<th>Age (years)</th>
<th>Level of laminectomy</th>
</tr>
</thead>
<tbody>
<tr>
<td>Specimen 01</td>
<td>Female</td>
<td>83</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 02</td>
<td>Female</td>
<td>78</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 03</td>
<td>Male</td>
<td>59</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 04</td>
<td>Female</td>
<td>84</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 05</td>
<td>Male</td>
<td>71</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 06</td>
<td>Male</td>
<td>88</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 07</td>
<td>Female</td>
<td>90</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 08</td>
<td>Male</td>
<td>79</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 09</td>
<td>Male</td>
<td>70</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 10</td>
<td>Male</td>
<td>65</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 11</td>
<td>Male</td>
<td>73</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 12</td>
<td>Female</td>
<td>83</td>
<td>L4</td>
</tr>
</tbody>
</table>

Table 2 gives an overview of the effects of laminectomy on the ROM. After single level laminectomy, the ROM increased significantly for FE (7.3%; $P = 0.018$), LB (7.5%; $P = 0.007$), and AR (12.2%; $P = 0.021$) in treated segments. No significant effects of laminectomy were found for control segments. ROM in LB of the adjacent level L3 – L4 decreased significantly (7.7%; $P = 0.033$) after laminectomy, while ROM of L3 – L4 levels in FE and AR remained unaffected.
Table 2. Statistical outcomes concerning spinal range of motion. The effect of laminectomy on the range of motion of spinal segments in flexion and extension, lateral bending and axial rotation. Effects of laminectomy on treated, adjacent and control segments were analyzed with repeated measures analysis of variance. SD is an abbreviation for standard deviation. Bold values indicate statistical significance at $P < 0.05$.

<table>
<thead>
<tr>
<th>Range of Motion</th>
<th>Segments</th>
<th>Untreated Degrees</th>
<th>After laminectomy Degrees</th>
<th>Untreated versus laminectomy</th>
<th>Segment level</th>
<th>Segment level x Laminectomy</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flexion and extension</td>
<td>Laminectomy 6x: L2–L3 and 5x: L4–L5</td>
<td>6.30 (SD: 2.44)</td>
<td>6.76 (SD: 2.58)</td>
<td>+7.26 0.018</td>
<td>0.322</td>
<td>0.265</td>
</tr>
<tr>
<td></td>
<td>Adjacent 12x: L3–L4</td>
<td>6.35 (SD: 2.55)</td>
<td>6.44 (SD: 2.39)</td>
<td>+1.45 0.656</td>
<td>0.825</td>
<td>0.903</td>
</tr>
<tr>
<td></td>
<td>Control 4x: L2–L3 and 6x: L4–L5</td>
<td>7.05 (SD: 2.14)</td>
<td>7.00 (SD: 2.17)</td>
<td>–0.73 0.967</td>
<td>0.236</td>
<td>0.392</td>
</tr>
<tr>
<td>Lateral bending</td>
<td>Laminectomy 6x: L2–L3 and 5x: L4–L5</td>
<td>4.82 (SD: 1.62)</td>
<td>5.18 (SD: 1.89)</td>
<td>+7.50 0.007</td>
<td>0.413</td>
<td>0.104</td>
</tr>
<tr>
<td></td>
<td>Adjacent 12x: L3–L4</td>
<td>5.73 (SD: 2.42)</td>
<td>5.28 (SD: 2.25)</td>
<td>–7.74 0.033</td>
<td>0.811</td>
<td>0.171</td>
</tr>
<tr>
<td></td>
<td>Control 4x: L2–L3 and 6x: L4–L5</td>
<td>5.87 (SD: 2.27)</td>
<td>5.81 (SD: 2.23)</td>
<td>–1.03 0.674</td>
<td>0.861</td>
<td>0.595</td>
</tr>
<tr>
<td>Axial rotation</td>
<td>Laminectomy 6x: L2–L3 and 5x: L4–L5</td>
<td>2.79 (SD: 1.90)</td>
<td>3.13 (SD: 2.06)</td>
<td>+12.19 0.021</td>
<td>0.616</td>
<td>0.896</td>
</tr>
<tr>
<td></td>
<td>Adjacent 12x: L3–L4</td>
<td>2.83 (SD: 1.62)</td>
<td>2.89 (SD: 1.68)</td>
<td>+2.20 0.537</td>
<td>0.803</td>
<td>0.241</td>
</tr>
<tr>
<td></td>
<td>Control 4x: L2–L3 and 6x: L4–L5</td>
<td>2.89 (SD: 1.49)</td>
<td>2.90 (SD: 1.37)</td>
<td>+0.29 0.839</td>
<td>0.315</td>
<td>0.334</td>
</tr>
</tbody>
</table>
Table 3. Statistical outcomes concerning spinal stiffness. The effect of laminectomy on the stiffness of spinal segments in flexion and extension, lateral bending and axial rotation. Effects of laminectomy on treated, adjacent and control segments were analyzed with repeated measures analysis of variance. SD is an abbreviation for standard deviation.

<table>
<thead>
<tr>
<th>Stiffness</th>
<th>Segments</th>
<th>Untreated</th>
<th>After laminectomy</th>
<th>Untreated versus laminectomy</th>
<th>Segment level</th>
<th>Segment level x Laminectomy</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Nm/Degree</td>
<td>Nm/Degree</td>
<td>%</td>
<td>P-values</td>
<td></td>
</tr>
<tr>
<td>Flexion and extension</td>
<td>Laminate</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>6x: L2–L3 and 5x: L4–L5</td>
<td>0.88 (SD: 0.75)</td>
<td>0.81 (SD: 0.68)</td>
<td>−8.34</td>
<td>0.126</td>
<td>0.740</td>
</tr>
<tr>
<td></td>
<td>Adjacent</td>
<td>12x: L3–L4</td>
<td>0.78 (SD: 0.60)</td>
<td>0.74 (SD: 0.45)</td>
<td>−5.22</td>
<td>0.527</td>
</tr>
<tr>
<td></td>
<td>Control</td>
<td>4x: L2–L3 and 6x: L4–L5</td>
<td>0.63 (SD: 0.26)</td>
<td>0.58 (SD: 0.28)</td>
<td>−6.65</td>
<td>0.354</td>
</tr>
<tr>
<td>Lateral bending</td>
<td>Laminate</td>
<td>6x: L2–L3 and 5x: L4–L5</td>
<td>1.28 (SD: 1.08)</td>
<td>1.19 (SD: 1.01)</td>
<td>−6.95</td>
<td>0.133</td>
</tr>
<tr>
<td></td>
<td>Adjacent</td>
<td>12x: L3–L4</td>
<td>1.25 (SD: 1.25)</td>
<td>1.11 (SD: 1.10)</td>
<td>−11.75</td>
<td>0.327</td>
</tr>
<tr>
<td></td>
<td>Control</td>
<td>4x: L2–L3 and 6x: L4–L5</td>
<td>1.03 (SD: 0.94)</td>
<td>0.96 (SD: 0.85)</td>
<td>−6.71</td>
<td>0.053</td>
</tr>
<tr>
<td>Axial rotation</td>
<td>Laminate</td>
<td>6x: L2–L3 and 5x: L4–L5</td>
<td>3.40 (SD: 3.08)</td>
<td>3.33 (SD: 3.01)</td>
<td>−1.98</td>
<td>0.880</td>
</tr>
<tr>
<td></td>
<td>Adjacent</td>
<td>12x: L3–L4</td>
<td>3.18 (SD: 3.33)</td>
<td>2.84 (SD: 2.59)</td>
<td>−10.72</td>
<td>0.174</td>
</tr>
<tr>
<td></td>
<td>Control</td>
<td>4x: L2–L3 and 6x: L4–L5</td>
<td>2.41 (SD: 1.43)</td>
<td>2.05 (SD: 0.94)</td>
<td>−14.96</td>
<td>0.058</td>
</tr>
</tbody>
</table>
Effects of laminectomy on stiffness are presented in Table 3. Laminectomy did not significantly affect stiffness of the treated segments (six times L2 – L3 and six times L4 – L5) in all three motion directions. Also, no significant effects of laminectomy were found for the stiffness of control segments (six times L2 – L3 and six times L4 – L5) and of adjacent segments (twelve times L3 – L4).

Finally, no significant effects for segment level and for the interaction between level and laminectomy were found for both ROM (Table 2) and stiffness (Table 3) in all motion directions.

**Discussion**

In the present study, we quantified the effects of facet-sparing single level laminectomy on the ROM and stiffness in FE, LB and AR under submaximal loading, using twelve fresh frozen human cadaveric lumbar spines. We found that laminectomy causes an increase in ROM of the treated segment while leaving stiffness unaffected. ROM and stiffness of adjacent segments also remained unaffected after laminectomy with exception of ROM in LB.

Other than previous studies, in which other types of uninstrumented and more extensive (multilevel) decompression surgery was performed in a similar test setup, this study also investigated the effect of decompression on adjacent segments. Previous studies found, in contrast to our results, a substantially larger increase in ROM at treated segment levels. Quint et al. (2008) showed an increase in ROM of 32 – 35 % during FE, 14 % during LB and 117 % during AR. However, these authors performed, besides a laminectomy, also a facetectomy and used a load-level of 7.5 Nm. Lee et al. (2010) also found substantially larger effects.

Although decompression techniques used by Lee et al. (2010) were similar to those in our study, differences in outcomes might have been caused by the application of higher loads (6 – 8 Nm) and axial compression (400 N) during testing. In the present study, a bending moment of 4 Nm was applied to the specimens in order to allow comparison with previous work and to anticipate on the fragility, and the subsequent risk of structural damage induced order effects of our relatively old and therefore possibly osteoporotic spines. It was previously established that osteoporotic specimens might be damaged during testing when high loads (around 7 – 8 Nm) are applied. We therefore restricted loading to 4 Nm, as even minimal damage to spinal structures might bias outcomes in subsequent testing procedures. Moreover, load deformation curves clearly leveled off at loads below 4 Nm (Figure 3).
A possible explanation for the significant effects of laminectomy on ROM, while leaving stiffness unaffected is that we measured stiffness between −1 Nm and +1 Nm. In this deformation zone, which basically represents the neutral zone, the spinous process, part of the lamina, and posterior attached ligaments are most likely either not strained or strained only within the toe region of this stress–strain curve and therefore significant effects of laminectomy are not found. At 4 Nm there may have been some deformation of these structures, which likely increased after laminectomy.

In our study, we found ROM in FE around 6 – 7 degrees, LB around 5 – 6 degrees and AR around 3 degrees. These ROMs are roughly 50% of the maximum range of motion that was found previously in vivo in healthy young adults 19, 30.

The neutral zone was previously hypothesized as a clinically relevant measure for instability of the lumbar spine 20, 21. Other studies determined the neutral zone as the zone in between the points of the largest changes in flexibility in the load–displacement curve 25. Unfortunately, these points could not reliably be detected in too many curves, as there were often small irregularities in load–displacement curves, possibly caused by degenerative deformities as a consequence of our aged sample. Consequently, we decided to measure stiffness between −1 Nm and +1 Nm in this study. As stated in the Introduction, a decrease in stiffness, as well as an increase in ROM, is from a biomechanical point-of-view the first sign of possible progression into an unstable situation and possibly increases the risk of progression into spondylolisthesis and/or spondylolysis.

Although spinal stiffness around the neutral zone remained unaffected after laminectomy, this type of decompression still results in a significantly increased ROM. Such an increase could lead to altered motion patterns and may therefore increase the risk of injury, including structural damage of the intervertebral disc and bony structures. However, when considering the magnitude of the effects on ROM in the present study, it seems plausible that these effects are not large enough to drastically increase the risk of clinical instability or failure at load levels investigated in the present study. Thus, the present findings do not seem to underline the urge for posterior instrumentation in order to (re–) stabilize the degenerative lumbar spine. However, previously we loaded single segments (L2 – L3 and L4 – L5) in axial rotation and shear loading under 1600 N compression 1, 5. Since these studies considered single spinal levels, we were able to apply a relatively high axial compressive load during testing in order to mimic physiological conditions 14, 15. Effects of laminectomy on stiffness were larger than in the present study, 20 % and 30 – 34 % for rotation and shear, respectively. Possibly, determination of stiffness in higher load regions, combined with the axial compression, caused these differences. In addition, strength parameters in these studies were equally affected by
laminectomy, a decrease of 18% after applying torsion moments and of a decrease of 44% in shear loading. These data suggest that considerations with regard to the question whether or not to apply posterior instrumentation at the treated and the adjacent level of the spine should possibly be based on failure risk \(^2, 3\) instead of on changes in ROM and stiffness around the neutral orientation. Note that we showed earlier that bone mineral density (BMD) is a strong predictor of the abovementioned failure forces \(^2, 3\).

Only a limited number of studies investigated the effects of decompression surgery on adjacent levels. Cardoso et al. (2008) \(^8\) found that adjacent instability occurs after a more extensive decompressive surgery (i.e. facetectomy). Similar to our results, Delank et al. (2010) \(^9\) found that adjacent levels were not substantially affected after decompressive surgery, even after facetectomy. Nevertheless, we did find a significant effect of laminectomy on ROM in LB (−7.8%; \(P = 0.033\)). Unfortunately, we are not able to explain these outcomes. A limitation of this study is that we combined both cranial (L3 – L4 with respect to laminectomy on L4) and distal adjacent (L3 – L4 with respect to laminectomy on L2) segments. The decision to analyze cranial and distal adjacent segments in a combined group was made on the basis of the limited number of specimen per group. We checked for effects of treated level in these segments and found no main effect or interaction. Therefore, we are confident that our results hold for both proximal and distal adjacent segments.

A possible confounder to this study is the effect of test sequence. For practical reasons, the order of testing was not completely randomized. While effects of test sequence cannot completely be excluded, we used more than one test sequence in order to limit bias. As we did not find any effects of repeated testing on our control segments, we can safely assume that sequence effects, if any, are small. Furthermore, we previously showed that repeated loading does have some small effects, but that these effects can be minimized by taking the tenth load cycle for ROM and by using an average over 10 cycles for stiffness \(^4\).

Another limitation is that we used a preload of 250 N for only 1 h. An axial preload can only partly simulate physiologic conditions, such as gravity and muscle forces. We did not apply axial loading during our test as application of compression combined with bending moments to a multi-segmented spine causes buckling \(^22\). Possibly, a short preload did not correspond with a daily loading pattern. Due to losses of fluids in the disc in daily life, the effect of laminectomy might be enhanced. A repetitive movement was performed to mimic repetitive loading strains and for consideration of visco- and poro-elastic behavior \(^4, 16, 31\). Furthermore, during daily in vivo loading, the lumbar spine is often subjected to a combination of different loading directions. Combined loading of the lumbar spine was not investigated in this study.
Finally, we only studied a commonly used type of laminectomy with preservation of the facet-joints in this study. Since it might be argued that more extensive types of decompression such as facetectomy can enhance, and less extensive types of decompression such as laminotomy can diminish the effects found in the present study, it is important to study these effects in future studies in a similar test setup to be able to allow for comparison with this specific study. Unfortunately, we could only use elderly but otherwise ‘healthy’ spines with no signs of spinal stenosis, which is normally the indication for laminectomy. However, we doubt that results would be much different had we used stenotic spines. The reference data for the untreated segments might be different, but these would then not reflect the normal spine biomechanics, which may be preferable as a reference to the effects of surgery. The treated segments would likely show the same kinematics as reported here, since structures causing the stenosis are removed by the laminectomy. Nevertheless, a follow-up study with stenotic spines would be useful, although this may not be feasible considering the limited availability of human cadaveric material.

In conclusion, we studied the ROM and stiffness around the neutral orientation of twelve human lumbar spines per segment (L2 – L3, L3 – L4 and L4 – L5) before and after single level facet-sparing laminectomy. We found that laminectomy significantly affects segmental ROM in all three motion directions. However, the magnitude of the increase is limited, i.e., between 7 % and 12 %. ROM of adjacent segments was only affected in LB. Stiffness of both treated and adjacent segments was not affected. The present results do not suggest that additional instrumentation would be needed as standard procedure to stabilize the spine when performing single level laminectomy. However, when the use of spinal instrumentation is considered, its primary focus should be on re-stabilizing the level at which the laminectomy was performed.
References


The effects of single level instrumented lumbar laminectomy on adjacent spinal biomechanics

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Abstract

**Purpose**
Posterior instrumentation is used to stabilize the spine after lumbar laminectomy. However, the effects on adjacent segmental stability are unknown. Therefore, we studied range of motion (ROM) and stiffness of treated lumbar spinal segments and cranial segments after laminectomy and after posterior instrumentation in flexion and extension (FE), lateral bending (LB) and axial rotation (AR). These outcomes might help understanding adjacent segment disease (ASD), which is reported cranial to posterior instrumentation.

**Methods**
Twelve cadaveric human lumbar spines were obtained. Spines were axially loaded with 250 N for one hour. Thereafter, ten consecutive load cycles (4 Nm) were applied in FE, LB, and AR. Subsequently, laminectomy was performed either at L2 or L4. Thereafter, load–deformation tests were repeated, after similar preloading. Finally, posterior instrumentation was added to the level treated with laminectomy before testing was repeated. The ROM and stiffness of the treated-, cranial adjacent- and control segments were calculated from load–displacement data. Repeated measures ANOVAs with level as between subject factor and laminectomy or instrumentation as within-subject factors were used.

**Results**
After laminectomy, ROM increased (+19.4 %) and stiffness decreased (–18.0 %) in AR. ROM in AR of adjacent segments was also increased (+11.0 %). ROM of treated segments after instrumentation decreased in FE (–74.3 %), LB (–71.6 %) and AR (–59.8 %). In adjacent segments after instrumentation, only ROM in LB was changed (–12.9 %).

**Conclusions**
The present findings do not substantiate a biomechanical pathway towards or explanation for ASD.
Introduction

When conservative treatment fails, symptomatic lumbar spinal stenosis is commonly treated with surgical decompression, often a laminectomy. Previously it has been shown in vitro that laminectomy reduces the threshold at which shear forces and torsion moments cause spondylolisthesis or spinal failure. Furthermore, it was also found that laminectomy alters biomechanical behavior under submaximal loading. In clinical practice, facet sparing laminectomy can cause spondylolisthesis with a reported incidence of 8 – 31 %. To prevent complications, surgeons often use posterior instrumentation, to obtain arthrodesis, i.e. fusion of a segment by bone formation.

However, rigid posterior instrumentation has several disadvantages associated with its use. The procedure of stabilization itself increases the probability of implant–related complications, including infection, pseudarthrosis, nerve injury, increased blood loss during surgery, extended surgery time and instrumentation failure. Furthermore, the use of instrumentation significantly increases the costs of surgery, especially since early re–operations are more common after laminectomy in combination with spinal fusion. Finally, the use of posterior instrumentation has been associated with adjacent segment disease (ASD). ASD is a symptomatic deterioration of the intervertebral disc adjacent to a previous fusion and is seen predominantly at cranial adjacent levels. Altered biomechanical behavior of the adjacent level, specifically an increased ROM and decreased stiffness as evidence of instability, due to the use of posterior instrumentation at a lower level might explain the development of ASD.

Previously, we found that adjacent biomechanical behavior was not substantially altered by laminectomy in flexion–extension (FE), lateral bending (LB) and axial rotation (AR). However, to our best knowledge no literature is available on the effects of instrumented single level facet–sparing laminectomy on cranial adjacent segments. Therefore, the aim of this study was to investigate the effects of instrumentation on the biomechanical behavior of adjacent segments in order to biomechanically substantiate the possible etiology of ASD cranial to posterior spinal instrumentation. Furthermore, we assessed to what extent posterior instrumentation after laminectomy actually restricts motion at the operated spine level.

Methods

Specimens and specimen preparation

Twelve human cadavers (mean age: 82.5 years, range 66 – 91) were included in this study. The bodies were donated to the department of anatomy by last will in accordance with the Dutch
The freshly frozen (−20 °C) cadavers were thawed before harvesting the lumbar spines (L1 – L5) and subsequent testing. Excessive soft tissue and muscle tissue was carefully removed, keeping the anterior and posterior longitudinal ligaments as well as the facet joints intact. During preparation, assessment and biomechanical testing, specimens were kept hydrated using 0.9 % saline–soaked gauzes. Anteroposterior, lateral and oblique radiographs (Sedical © Digital Vet. DX–6, Arlington Heights, IL, USA) were made to determine whether bridging osteophytes were present in segments. Lumbar spines with bridging osteophytes were excluded from this study. Magnetic resonance imaging (MRI, Siemens © Symphony 1.5 T: Syngo MR A30, software NUMARIS/4, Berlin, Germany), lateral and oblique radiographs and visual inspection also confirmed that facet joints were intact and no fractures of the pars interarticularis were present in segments before mechanical testing.

After imaging, the top and bottom vertebrae (L1 and L5) were potted in a casting–mold and partially buried in a low melting point (48 °C) bismuth alloy (Cerrolow – 147; 48.0 % bismuth, 25.6 % lead, 12.0 % tin, 9.6 % cadmium, and 4.0 % indium) (Figure 1). The L1 and L5 vertebral bodies were firmly fixed into the alloy by adding screws into the vertebral body prior to submerging in the alloy. Discs were placed parallel based on visual inspection. Because muscle tissue was thoroughly and carefully removed, the intervertebral disc and corresponding endplates were clearly visible. All articulating parts were kept free. Markers containing 3 LEDs were rigidly fixed with screws to the casting-mold containing L1, the anterior surface of the vertebral bodies of L2, L3 and L4 and to the casting–mold in which L5 was mounted.

**Biomechanical testing**

The test setup was similar to previous studies 12-15. Lumbar spines were placed horizontally in a custom made four–points bending device in which FE, LB and AR was applied using a hydraulic materials testing machine (Instron ©, model 8872; Instron and IST, Norwood, Canada). This setup guarantees equal moments at all levels of the lumbar spine.

Before testing, a compressive preload of 250 N was applied for one hour. A pure axial compressive force was applied using a pneumatic cylinder. Calibration of axial compression was performed using a load cell (Houthinger Baldwin Messtechnik ©, Force Transducer Type C2, Darmstadt, Germany). The chosen amount of axial preload was selected to allow for comparison with previous work 12-15 and to mimic physiological conditions. During testing, no compressive load was applied in order to prevent buckling of the multi–segmented spine 16. Loads were applied up to a moment of 4 Nm at a constant angular velocity of 0.5 degree/
When a moment of +4 Nm was measured, the Instron reversed its loading direction until –4 Nm was reached. Each movement direction was tested for ten consecutive cycles. Force and displacement of the Instron were recorded and digitized at 100 Hz (Instron © Fast Track 2). All tests were performed at room temperature.

During testing, motions of the LEDs of the casting-mold containing L1, on the vertebral bodies of L2, L3, L4 and the casting–mold in which L5 was mounted were recorded by an optoelectronic 3–dimensional movement registration system with 1 array of 3 cameras (Optotrak © 3020, Northern Digital Inc., Waterloo, ON). The 3–dimensional resolution of this system at a distance of 2 m is 0.01 mm. Before testing, the axes of the Optotrak were aligned with the anatomic axes of the spines. Labview software was used for data acquisition, synchronized with Instron data. The sampling rate was 100 samples per second.

After the first set of measurements (FE, LB and AR), laminectomy was performed at level L2 of six randomly chosen lumbar spines, and at level L4 of the remaining six lumbar spines. The cranial adjacent levels (six times segment L1 – L2 and six times segment L3 – L4) were studied as well as the more distant untreated segments (six times segment L4 – L5 and six times segment L2 – L3), which were used as control group. Laminectomy, analogous to standard clinical practice, was performed by removing the spinous process and part of the lamina, leaving the facet joints intact (Figure 1). Again, a compressive preload of 250 N was applied for one hour. Thereafter lumbar spines were tested again in AR, LB and FE for another set of ten consecutive cycles. Before the last series of tests, standard posterior lumbar fusion (PLF) was performed by placing pedicle screws and rods (Medtronic ©, CD Horizon Legacy Spinal System, Minneapolis, MN, United States) at the level of the laminectomy (Figure 1). Placement of screws was assessed on anterior–posterior and lateral radiographs to confirm the correct position in the pedicle. No interbody devices were used.

The first six lumbar spines (three times laminectomy on L2 and three times laminectomy on L4) were tested in order A: FE–LB–AR–laminectomy–AR–LB–FE–instrumentation–FE–LB–AR while the second six lumbar spines (three times laminectomy on L2 and three times laminectomy on L4) were tested in order B: AR–FE–LB–laminectomy–LB–FE–AR–instrumentation–AR–FE–LB. The order of testing was changed to correct for possible effects of test sequence.

**Data analysis**

Using Instron forces, the dimension of the 4-point bending device and Optotrak LED displacements a Matlab (Mathworks ©, Natick, MA, USA) computer program calculated the load–displacement curves in the loaded direction for L2 relative to L1, for L3 relative to L2,
for L4 relative to L3 and for L5 relative to L4. For each individual test (FE, LB and AR), the ROM (degrees) and stiffness (Nm/degree) of untreated spines, and spines after single level laminectomy and after subsequent instrumentation were calculated per motion segment (L1 – L2, L2 – L3, L3 – L4 and L4 – L5) from the load–displacement data. The ROM was calculated between an applied load of –4 Nm and +4 Nm. The tenth cycle was used for analysis 12. Stiffness was estimated by means of a least squares fit of a straight line through a fitted curve of the load–displacement data with the slope of the fitted line representing stiffness. Stiffness was calculated between – 1.0 Nm and + 1.0 Nm.

Figure 1. Biomechanical testing sequence with an untreated lumbar spine (on the left), a lumbar spine after laminectomy (in the middle) and a lumbar spine after laminectomy and posterior instrumentation (on the right). Note that laminectomy was performed on L4 (as shown in this figure) in six segments, while in the other six segments (instrumented) laminectomy was performed on level L2.
**Statistical methods**

Effects of laminectomy and instrumentation on ROM and stiffness were tested using repeated measures Analysis of Variance (ANOVA) with level as between subject factor and laminectomy or instrumentation as within subject factor. Levels L2 – L3 and L4 – L5 were considered as intervention group with respect to laminectomy at L2 and L4, respectively. Levels L1 – L2 and L3 – L4 were considered as cranial adjacent for laminectomy at L2 and L4, respectively. Control groups consisted of L2 – L3 and L4 – L5 with respect to laminectomy at L4 and L2, respectively.

First, ROM and stiffness in FE, LB and AR of treated, adjacent and control segments before and after laminectomy were compared. Secondly, ROM and stiffness in FE, LB and AR of treated, adjacent and control segments before and after instrumented laminectomy were compared. A significance level of 5 % was used. The statistical analyses were performed using SPSS for Mac version 20.0 (SPSS Inc. ©, Chicago, IL, USA).

**Results**

An overview of specimen characteristics is presented in Table 1. Tables 2 and 3 provide an overview of the effects of laminectomy and instrumentation on the ROM and stiffness, respectively. Effects of segment level and interaction effects with treatment are presented in Table 4.

ROM and stiffness in all motion directions of specimen number 02 (L2 – L3 and L3 – L4) and specimen number 12 (L3 – L4) were excluded from analysis, due to severely irregular load–displacement curves, leaving 45 segments to be analyzed. Data of segments L3 – L4 and L4 – L5 of specimen number 03 after instrumentation had to be excluded due to a loosened marker attached to the vertebral body of L4. Both ROM and stiffness data of these measurements (FE, LB and AR) were excluded. Due to a limited ROM of treated segments after instrumentation, the resultant load-displacement curves were too irregular to be able to accurately define stiffness. Therefore we did not calculate stiffness at the instrumented level (as shown with an ‘X’ in Tables 3 and 4). Furthermore, the resultant range of ROM was too small (<1 degree) in 22 measurements to allow for reliable assessment of stiffness, which were therefore excluded for final analysis. In a few cases, not all ten cycles were used due to irregular curves or other errors related to calculation.
Table 1. Overview of specimens

<table>
<thead>
<tr>
<th>Donor</th>
<th>Sex</th>
<th>Age</th>
<th>Laminectomy</th>
</tr>
</thead>
<tbody>
<tr>
<td>Specimen 01</td>
<td>Female</td>
<td>83</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 02</td>
<td>Male</td>
<td>76</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 03</td>
<td>Female</td>
<td>91</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 04</td>
<td>Female</td>
<td>86</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 05</td>
<td>Male</td>
<td>82</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 06</td>
<td>Male</td>
<td>82</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 07</td>
<td>Male</td>
<td>66</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 08</td>
<td>Male</td>
<td>72</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 09</td>
<td>Male</td>
<td>90</td>
<td>L2</td>
</tr>
<tr>
<td>Specimen 10</td>
<td>Male</td>
<td>86</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 11</td>
<td>Female</td>
<td>86</td>
<td>L4</td>
</tr>
<tr>
<td>Specimen 12</td>
<td>Male</td>
<td>90</td>
<td>L2</td>
</tr>
</tbody>
</table>

Spinal segments after laminectomy and after instrumented laminectomy (L2 – L3 and L4 – L5)

ROM increased significantly for AR (+19.4 %; \( P = 0.001 \)) in segments treated with a laminectomy, while no significant effects were observed for FE and LB. As for ROM, stiffness after laminectomy was only affected in AR (-18.0 %; \( P = 0.005 \)). Again, no significant effects were found for FE and LB. After instrumentation, ROM decreased significantly in FE (-74.3 %; \( P = 0.001 \)), LB (-71.6 %; \( P = 0.001 \)) and AR (-59.8 %; \( P < 0.001 \)), compared to after laminectomy. Stiffness of instrumented segments was, as stated, not calculated.

Adjacent segments (L1–L2 and L3–L4)

ROM of adjacent segments in AR increased significantly (+11.0 %; \( P = 0.043 \)) after laminectomy. No significant effects on ROM in FE and LB of the adjacent segments were observed. After laminectomy, no significant effects on spinal stiffness of adjacent segments in any of the three directions were found. Instrumentation caused a significant decrease of ROM of adjacent segments in LB (-12.9 %; \( P = 0.025 \)), compared to after laminectomy. No significant effects on ROM in FE and AR were found. Stiffness of segments adjacent to instrumented segments was not affected in FE, LB and AR.

Control segments (L2 – L3 and L4 – L5)

After laminectomy, ROM in FE of the control segments increased significantly (\( P = 0.043 \)) by 8.6 %. In addition, stiffness of control segments in LB was significantly decreased (-12.2 %; \( P = 0.006 \)). After instrumentation, ROM and stiffness of control segments were not different from after laminectomy.
Table 2. The effect of laminectomy and instrumentation on the range of motion (ROM) of spinal segments in flexion and extension (FE), lateral bending (LB) and axial rotation (AR). Effects of laminectomy and instrumentation on treated, adjacent and control segments were analyzed with ANOVA. In case a measurement was excluded, the ANOVA also excludes its corresponding value. These exclusions of data are shown with ‘n=’ behind each value.

<table>
<thead>
<tr>
<th>Range of Motion</th>
<th>Segments</th>
<th>Untreated Degrees</th>
<th>After laminectomy Degrees</th>
<th>After instrumentation Degrees</th>
<th>Untreated versus laminectomy %</th>
<th>Untreated versus instrumentation P-value</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>FE Laminectomy</td>
<td>5x: L2–L3 and 6x: L4–L5</td>
<td>5.28 (± 2.28) (n=11)</td>
<td>5.81 (± 2.76) (n=11)</td>
<td>1.41 (± 0.90) (n=10)</td>
<td>+10.1</td>
<td>0.106 (n=11)</td>
<td>–74.3</td>
</tr>
<tr>
<td>Adjacent</td>
<td>6x: L1–L2 and 6x: L3–L4</td>
<td>4.42 (± 1.70) (n=12)</td>
<td>4.46 (± 1.74) (n=12)</td>
<td>4.85 (± 1.70) (n=11)</td>
<td>+0.9</td>
<td>0.863 (n=12)</td>
<td>+8.2</td>
</tr>
<tr>
<td>Control</td>
<td>6x: L2–L3 and 6x: L4–L5</td>
<td>4.51 (± 1.94) (n=12)</td>
<td>4.90 (± 2.10) (n=12)</td>
<td>4.25 (± 1.82) (n=12)</td>
<td>+8.6</td>
<td>0.043 (n=12)</td>
<td>–13.2</td>
</tr>
<tr>
<td>LB Laminectomy</td>
<td>5x: L2–L3 and 6x: L4–L5</td>
<td>4.87 (± 2.31) (n=11)</td>
<td>4.96 (± 2.28) (n=11)</td>
<td>1.33 (± 0.97) (n=10)</td>
<td>+1.9</td>
<td>0.416 (n=11)</td>
<td>–71.6</td>
</tr>
<tr>
<td>Adjacent</td>
<td>6x: L1–L2 and 6x: L3–L4</td>
<td>4.00 (± 2.22) (n=12)</td>
<td>4.32 (± 2.41) (n=12)</td>
<td>3.86 (± 1.82) (n=11)</td>
<td>+8.1</td>
<td>0.076 (n=12)</td>
<td>–12.9</td>
</tr>
<tr>
<td>Control</td>
<td>6x: L2–L3 and 6x: L4–L5</td>
<td>3.86 (± 2.40) (n=12)</td>
<td>4.37 (± 2.65) (n=12)</td>
<td>4.10 (± 2.58) (n=12)</td>
<td>+13.4</td>
<td>0.073 (n=12)</td>
<td>–6.2</td>
</tr>
<tr>
<td>AR Laminectomy</td>
<td>5x: L2–L3 and 6x: L4–L5</td>
<td>2.65 (± 1.09) (n=11)</td>
<td>3.17 (± 1.19) (n=11)</td>
<td>1.32 (± 0.57) (n=10)</td>
<td>+19.4</td>
<td>0.001 (n=11)</td>
<td>–59.8</td>
</tr>
<tr>
<td>Adjacent</td>
<td>6x: L1–L2 and 6x: L3–L4</td>
<td>2.13 (± 1.43) (n=12)</td>
<td>2.37 (± 1.48) (n=12)</td>
<td>2.18 (± 1.49) (n=11)</td>
<td>+11.0</td>
<td>0.043 (n=12)</td>
<td>–8.9</td>
</tr>
<tr>
<td>Control</td>
<td>6x: L2–L3 and 6x: L4–L5</td>
<td>2.58 (± 1.61) (n=12)</td>
<td>2.69 (± 1.78) (n=12)</td>
<td>2.65 (± 1.64) (n=12)</td>
<td>+4.3</td>
<td>0.096 (n=12)</td>
<td>–1.3</td>
</tr>
</tbody>
</table>
Table 3. The effect of laminectomy and instrumentation on the stiffness of spinal segments in flexion and extension (FE), lateral bending (LB) and axial rotation (AR). Effects of laminectomy and instrumentation on treated, adjacent and control segments were analyzed with ANOVA. Spinal stiffness at the level of instrumentation was not analyzed, which is shown by an ‘X’. In case a measurement was excluded, the ANOVA also excludes its corresponding value. These exclusions of data are shown with ‘n=’ behind each value.

<table>
<thead>
<tr>
<th>Stiffness</th>
<th>Segments</th>
<th>Untreated</th>
<th>After laminectomy</th>
<th>After instrumentation</th>
<th>Untreated versus laminectomy</th>
<th>Laminectomy versus instrumentation</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Degrees</td>
<td>Degrees</td>
<td>Degrees</td>
<td>%</td>
<td>P-value</td>
</tr>
<tr>
<td>FE</td>
<td>Laminectomy 5x: L2–L3 and 6x: L4–L5</td>
<td>0.81 (± 0.44) (n=9)</td>
<td>0.70 (± 0.35) (n=9)</td>
<td>X</td>
<td>−14.4</td>
<td>0.129 (n=9)</td>
</tr>
<tr>
<td></td>
<td>Adjacent 6x: L1–L2 and 6x: L3–L4</td>
<td>1.56 (± 1.20) (n=12)</td>
<td>1.33 (± 1.10) (n=12)</td>
<td>1.24 (± 0.68) (n=11)</td>
<td>−14.7</td>
<td>0.080 (n=12)</td>
</tr>
<tr>
<td></td>
<td>Control 6x: L2–L3 and 6x: L4–L5</td>
<td>1.14 (± 0.58) (n=12)</td>
<td>1.10 (± 0.56) (n=12)</td>
<td>1.14 (± 0.56) (n=11)</td>
<td>−4.3</td>
<td>0.417 (n=12)</td>
</tr>
<tr>
<td>LB</td>
<td>Laminectomy 5x: L2–L3 and 6x: L4–L5</td>
<td>1.49 (± 1.91) (n=9)</td>
<td>1.29 (± 1.24) (n=9)</td>
<td>X</td>
<td>−13.4</td>
<td>0.604 (n=9)</td>
</tr>
<tr>
<td></td>
<td>Adjacent 6x: L1–L2 and 6x: L3–L4</td>
<td>1.73 (± 1.32) (n=12)</td>
<td>1.50 (± 0.91) (n=12)</td>
<td>1.87 (± 1.52) (n=11)</td>
<td>−13.6</td>
<td>0.231 (n=12)</td>
</tr>
<tr>
<td></td>
<td>Control 6x: L2–L3 and 6x: L4–L5</td>
<td>1.90 (± 1.57) (n=12)</td>
<td>1.67 (± 1.43) (n=12)</td>
<td>1.63 (± 1.00) (n=11)</td>
<td>−12.2</td>
<td>0.006 (n=12)</td>
</tr>
<tr>
<td>AR</td>
<td>Laminectomy 5x: L2–L3 and 6x: L4–L5</td>
<td>2.63 (± 1.58) (n=11)</td>
<td>2.16 (± 1.33) (n=11)</td>
<td>X</td>
<td>−18.0</td>
<td>0.005 (n=11)</td>
</tr>
<tr>
<td></td>
<td>Adjacent 6x: L1–L2 and 6x: L3–L4</td>
<td>2.43 (± 1.34) (n=9)</td>
<td>2.16 (± 1.21) (n=9)</td>
<td>2.35 (± 1.44) (n=8)</td>
<td>−10.8</td>
<td>0.533 (n=9)</td>
</tr>
<tr>
<td></td>
<td>Control 6x: L2–L3 and 6x: L4–L5</td>
<td>2.42 (± 1.13) (n=11)</td>
<td>2.25 (± 1.16) (n=11)</td>
<td>2.40 (± 1.08) (n=11)</td>
<td>−6.8</td>
<td>0.177 (n=11)</td>
</tr>
</tbody>
</table>
Table 4. Effects of level, and interaction effects level x laminectomy and level x instrumentation on ROM and stiffness. Spinal stiffness at the level of instrumentation was not analyzed, which is shown by an ‘X’.

<table>
<thead>
<tr>
<th></th>
<th>Segments</th>
<th>Segment level</th>
<th>Segment level x Laminectomy</th>
<th>Segment level x Instrumentation</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>ROM</td>
<td>Stiffness</td>
<td>ROM</td>
</tr>
<tr>
<td>FE</td>
<td>Laminectomy</td>
<td>0.414</td>
<td>0.665</td>
<td>0.255</td>
</tr>
<tr>
<td></td>
<td>Adjacent</td>
<td>0.721</td>
<td>0.448</td>
<td>0.516</td>
</tr>
<tr>
<td></td>
<td>Control</td>
<td>0.707</td>
<td>0.626</td>
<td>0.535</td>
</tr>
<tr>
<td>LB</td>
<td>Laminectomy</td>
<td>0.441</td>
<td>0.780</td>
<td>0.151</td>
</tr>
<tr>
<td></td>
<td>Adjacent</td>
<td>0.576</td>
<td>0.291</td>
<td>0.178</td>
</tr>
<tr>
<td></td>
<td>Control</td>
<td>0.917</td>
<td>0.337</td>
<td>0.368</td>
</tr>
<tr>
<td>AR</td>
<td>Laminectomy</td>
<td>0.563</td>
<td>0.987</td>
<td>0.249</td>
</tr>
<tr>
<td></td>
<td>Adjacent</td>
<td>0.108</td>
<td>0.826</td>
<td>0.384</td>
</tr>
<tr>
<td></td>
<td>Control</td>
<td>0.416</td>
<td>0.550 (0.032)</td>
<td>0.387</td>
</tr>
</tbody>
</table>
**Effects of segment level**

Finally, Table 4 shows that, besides for the interaction between segment level and laminectomy in the control group of AR ($P = 0.039$) no significant effects of segment level or interactions between treatment (both laminectomy and instrumentation) and segment level were found.

**Discussion**

In the present study, we quantified the effects of single level facet sparing laminectomy and posterior instrumentation on the ROM and stiffness of adjacent spinal segments in twelve fresh frozen human cadaveric lumbar spines. We found that laminectomy increases the ROM of both treated and adjacent segments in AR, while AR stiffness was only decreased at the treated level. As expected, posterior instrumentation significantly and substantially decreased the ROM in FE, LB and AR, while in adjacent segments only a decrease in LB ROM was found. Spinal stiffness of adjacent segments was unaffected by instrumentation.

Previously, we studied the effects of uninstrumented laminectomy on adjacent biomechanics. In that study we found a significant increase in ROM at the treated level after laminectomy in FE, LB and AR, ranging from $7–12\%$. Although our current results only statistically confirmed this effect of laminectomy for AR ($+19.4\%$; $P = 0.001$), results in FE were comparable but not significant ($+10.1\%$; $P = 0.106$), while LB results did not show this increase ($+1.9\%$; $P = 0.416$). Differences between these studies can possibly be explained by differences in state of degeneration between studies. Interestingly, in both the present and previous study, effects of laminectomy were most substantial for AR. A possible explanation is that a laminectomy disrupts the integrity of the posterior arch, while AR moments are largely resisted by the arch whereas the intervertebral disc contributes substantially to resistance against FE and LB. Similar effects of laminectomy on AR were previously described. The fact that the posterior arches of segments L1 – L2 and L3 – L4 level are supported by those of the treated segments (L2 – L3 and L4 – L5) could cause a laminectomy at level L2 or L4 to also increase ROM and decrease stiffness at the cranial adjacent segment, albeit to a lesser extent. While changes in average values in the present data do suggest such effects to some extent, these effects were only found to be statistically significant for ROM in AR.

In previous work, we elaborated on the question why significant alterations in ROM do not always coincide with significant alterations in corresponding stiffness. We believe, since we measured stiffness between $-1Nm$ and $+1Nm$, which basically represents the neutral zone, the spinous process, part of the lamina, and posterior attached ligaments are most likely either not strained or strained only within the toe region of their stress-strain curves. At $4Nm$
these structures would be strained more substantially, due to which they would contribute more to the resistance against movement.

New in this study was the investigation of effects of a single level lumbar laminectomy and additional instrumentation on adjacent segments. To our best knowledge, only Cardoso et al. and Delank et al. investigated effects of laminectomy and instrumentation on adjacent levels. However, these authors used different types of (multilevel) decompressive techniques, such as facetectomy and performed other (multilevel) constructs, making it difficult to compare results.

We found that, as expected, instrumentation substantially reduces ROM at the level at which instrumentation is used by as much as 60 – 74 %. In addition, instrumentation slightly but significantly decreased ROM of the adjacent segment in lateral bending by 13 %, while FE and AR ROM and stiffness in all directions was unaffected. This result does not seem to offer a biomechanical explanation for ASD. However, in this study we did not investigate the effect of posterior lumbar interbody fusion (PLIF) on adjacent levels. It is progressively becoming clear that, in particular, the use of PLIF and PLF holds a risk of ASD, which is receiving increasing interest among spine surgeons. ASD is one of the reasons why re–operations are necessary and has a higher incidence in elderly patients. The probability of undergoing revision surgery for ASD was 5.8 % at five years and 10.4 % and ten years postoperatively. The prevalence of ASD requiring re–operation in patients older than 60 years of age was 21.9 % at ten years postoperatively. PLIF procedures showed significantly lower survival rates than PLF procedures. ASD is often found at the level cranial to spinal fusion. The exact pathogenesis is not well understood. First, it has been argued that ASD merely reflects the progression of physiological degeneration to other than the primarily affected segment. Secondly, it has been suggested that application of instrumentation could negatively affect spinal stability of the adjacent level, which could speed up the process of degeneration. The first hypothesis does not explain why mainly the cranial adjacent segment progresses into a state of degeneration. The second hypothesis is not supported by the biomechanical results presented here. If subjects would, in daily life, impose the same amount of total lumbar motion, as before surgery, it is clear that after instrumentation, the ROM of adjacent segments would increase. However, this still does not explain the predominance of ASD in segments cranial to the treated segment.

In clinical practice, topping–off procedures are used with the aim of preventing or slowing down ASD. Topping–off procedures combine rigid fusion with a flexible pedicle screw system at the cranial adjacent level to prevent ASD. In our opinion, these procedures would only be beneficial when stability is negatively affected at the adjacent level. However, in contrast,
we found that spinal ROM at the adjacent level after instrumentation was decreased, while instability would coincide with an increase in ROM. Apparently, instrumentation does not only stabilize the treated segment, it also supports the posterior arch of the lower level of the adjacent segment. Therefore, our results do not support the use of topping-off procedures to re-stabilize adjacent levels after posterior instrumentation.

One limitation of the present study is that we were not able to evaluate distal adjacent levels. The segment distal to L4 – L5 was not included as L5 was our bottom vertebra. While data of the segments distal to L2 – L3 were available, these included only 6 segments, of which 2 had to be excluded due to irregular load-deformation curves. Nevertheless, in clinical practice the biomechanical behaviour of the cranial adjacent segments has more relevance.

During analysis, we found three unexpected significant results. ROM in FE of the control segments increased significantly \((P = 0.043)\) by 8.6 %. Stiffness in LB of the control segments decreased significantly \((P = 0.006)\) after laminectomy with 12.2 %. These effects may indicate tissue creep due to repeated testing or dehydration, which would suggest that we slightly overestimated ROM and underestimated stiffness after laminectomy and even more so after instrumentation. In addition, a significant \((P = 0.039)\) interaction between segment level and laminectomy in the control group of AR was found, for which we can offer no explanation.

Stiffness was calculated in the neutral zone as this been hypothesized to be a clinically relevant indicator of instability of the lumbar spine. \(^{26, 27}\) Other studies determined the neutral zone as the zone in between the points of the largest changes in flexibility in the load–displacement curve. \(^{28}\) Unfortunately, these points could not reliably be detected in too many curves, as there were often small irregularities in load-displacement curves, possibly caused by degenerative deformities as a consequence of our aged sample. Consequently, we decided to measure stiffness between −1 Nm and +1 Nm in this study.

The one hour, 250 N axial preload applied can only partly simulate effects of loading due to gravity and muscle forces. We did not apply axial loading during our test as application of compression combined with bending moments to a multi–segmented spine causes buckling. \(^{16}\) Possibly, the short preload period did not correspond with a daily loading pattern. Due to losses of fluids in the disc in daily life, the effects of laminectomy might be enhanced. We repeated each load cycle 10 times, thereby allowing for some visco– and poro– elastic behavior to mimic \textit{in vivo} loading. \(^{12, 29, 30}\) Furthermore, during daily \textit{in vivo} loading, the lumbar spine is often subjected to a combination of different loading directions. Combined loading of the lumbar spine was not investigated in this study.
In conclusion, we found that, while posterior instrumentation does stabilize segments treated with a laminectomy, it does not negatively alter (i.e., increase ROM or decrease stiffness) adjacent spinal biomechanics. Based on our results, spinal segments cranial to posterior fusion techniques do not become less stable after instrumentation. Therefore, we postulate that altered biomechanical behavior due laminectomy with or without stabilization does not lead substantial alterations at the adjacent level and could therefore not be the explanation for ASD.
References


9

General discussion
The present thesis concerns an in depth biomechanical analysis related to single level lumbar laminectomy. In orthopedic and neurosurgical practice, elderly patients often present with symptomatic degenerative lumbar spinal stenosis, which can be treated with a facet-sparing laminectomy. Post-operative symptomatic clinical instability, which is considered as a serious complication, justifies re-operation to stabilize and fuse the unstable segment. Up to now, there are no clear criteria that define the need for additional instrumentation following a single level lumbar laminectomy. Although spinal stability increases after the application of posterior instrumentation, the procedure of stabilization itself increases the probability of implant-related complications, including adjacent segment degeneration (ASD). Moreover, it significantly increases the costs of patient care. Furthermore, back pain and complaints related to spinal pathology are among the most common health problems and are associated with high healthcare costs and productivity losses. Due to the ageing population, these costs can be expected to increase in the next decades.

This thesis is the first to provide a complete overview of human cadaveric spinal biomechanics before and after single level lumbar laminectomy in the aged lumbar spine and the biomechanical effects of additional posterior instrumentation. In case a single level lumbar laminectomy was biomechanically evaluated in literature, it usually considered an animal or finite element model. In addition, more extensive decompressive techniques were previously published, while adjacent segment instability was only investigated in a limited number of studies. Literature describing the biomechanical effects of adjacent segments after rigid posterior instrumentation is also scarce.

As introduced in Chapter 1, this comprehensive biomechanical study using human lumbar spines considers the investigation of three main questions. First, how does single level laminectomy affect spinal ROM, stiffness and strength? Secondly, can spinal instability following lumbar laminectomy be prognosticated based on clinically measurable imaging parameters? Thirdly, what are the biomechanical effects under submaximal loading conditions of single level laminectomy on the treated and adjacent levels with and without instrumentation of the treated segment?

To answer these main questions, a total of seven experimental sub questions were identified and studied in a similar number of studies (Chapters 2 – 8). Chapters 2 – 5 considered single level (L2 – L3 or L4 – L5) testing experiments until failure, while chapters 6 – 8 considered submaximal testing of the whole lumbar spine (L1 – L5). In the latter, stiffness was calculated around the neutral zone. All experimental studies were performed on custom-made testing machines by using fresh frozen human lumbar cadaveric spines, thereby enhancing the clinical relevance of our data. All seven study aims, as presented in the introduction, were met by these studies.
1. To assess the effects of single level lumbar laminectomy on shear biomechanics of the treated segment (Chapter 2).
2. To assess how shear biomechanics after single level lumbar laminectomy can be predicted (Chapter 3).
3. To assess the effects of single level lumbar laminectomy on torsion biomechanics of the treated segment (Chapter 4).
4. To assess how torsion biomechanics after single level lumbar laminectomy can be predicted (Chapter 5).
5. To assess the methodology for submaximal biomechanical testing of lumbar spinal segments (Chapter 6).
6. To quantify the effects of laminectomy on submaximal biomechanical behaviour of treated and adjacent lumbar segments (Chapter 7).
7. To quantify the effects of instrumented laminectomy on submaximal biomechanical behaviour of treated and adjacent lumbar segments (Chapter 8).

Shear biomechanics after laminectomy: effects and prediction on stability
The 2nd and 3rd chapter showed that laminectomy results in a decrease of shear force to failure (SFF) (44.2 %), shear displacement at failure (DF) (38.6 %) and shear stiffness (SS) (19.9 %). For segments with laminectomy, SS was significantly correlated with intervertebral disc degeneration and facet joint degeneration (Pfirrmann, Griffith, Lane and Pathria). Shear yield force (SYF) as a measure of the transition from elastic to plastic deformation, representing the first damage to a structure decreased 41.1 % after laminectomy and was correlated with intervertebral disc geometry (length, surface and volume), bone mineral content (BMC) and frontal area. SFF was correlated with disc length, BMC and bone mineral density (BMD). Using selections of the above-mentioned variables, SS, SYF and SFF could reasonably well to well be predicted for segments with laminectomy ($r^2$-values respectively: 0.53, 0.81 and 0.77). In conclusion, significant loss of strength and SS were predicted by DXA-derived measures of bone quantity and quality (respectively, BMC and BMD), intervertebral disc geometry and degenerative parameters, suggesting that low BMC or BMD, small intervertebral discs and absence of osteophytes could predict the possible development of postoperative shear instability following lumbar laminectomy.

Torsion biomechanics after laminectomy: effects and prediction on stability
In chapters 4 and 5 it was found that load–displacement curves have a typical bi-phasic pattern with a distinct early torsion stiffness (ETS) and late torsion stiffness (LTS). Following laminectomy, a significant and substantial decrease in ETS (34.1 %) and LTS (30.1 %) was found, whereas the torsion moment to failure (TMF) decreased to a lesser extent, with 17.6
%. Univariate analyses showed that a range of geometric characteristics and disc and bone quality parameters were also associated with torsion biomechanical properties of lumbar segments. Multivariate models showed that ETS, LTS and TMF could be predicted very well for segments after laminectomy ($r^2$-values 0.95, 0.87 and 0.93, respectively), with, just like for shear biomechanics, BMC and BMD being the main predictors. Furthermore, geometry parameters, i.e. intervertebral disc width, frontal area and facet joint tropism, were found to be predictors of ETS, LTS and TMF following laminectomy. In conclusion, as for shear biomechanics, vertebral bone content and geometry parameters can be used to predict the possible development of postoperative rotational instability following lumbar laminectomy.

**Methodological assessment of submaximal testing of spinal biomechanics**

The 6th chapter presents a methodological paper concerning the current gold standard in submaximal testing procedures for the lumbar spine in flexion-extension (FE), lateral bending (LB) and axial rotation (AR). It was found that the ROM increased slightly but significantly in all directions after three (FE: 1.0 %, LB: 1.5 %, and AR: 1.5 %) and after ten loading cycles (FE: 2.9 %, LB: 3.3 %, and AR: 3.3 %). Stiffness, measured around the neutral zone, was not significantly affected, but varied considerably over cycles. It was concluded that, although effects were small, assessment of the tenth cycle instead of the regularly used third cycle reduces viscoelastic effects in repeated measurements of ROM, because the spine is closer to a steady state condition. Furthermore, averaging over loading cycles would improve the assessment of stiffness estimates.

**Effects of laminectomy on submaximal behaviour of treated and adjacent segments**

Chapter 7 showed that, in submaximal testing, ROM at the level of laminectomy increased significantly after laminectomy for FE (7.3 %), LB (7.5 %), and AR (12.2 %). ROM of adjacent segments was not affected significantly, with exception of LB (– 7.7 %). Control segments were not affected. Spinal stiffness of treated, adjacent and control segments was not affected by laminectomy.

**Effects of instrumented laminectomy on submaximal behaviour of treated and adjacent segments**

In chapter 8, we found that ROM in AR increased (19.4 %) and stiffness decreased (– 18.0 %) significantly after laminectomy in treated segments, while FE and LB were not significantly affected. After laminectomy, ROM in AR of adjacent segments was also increased (11.0 %). Again FE and LB remained unaffected. Thus, as for chapter 7, laminectomy had limited effects on submaximal biomechanics of the treated and had only minor effects on adjacent segments. Instrumentation, however, significantly decreased ROM of treated segments in FE (– 74.3 %), LB (– 71.6 %) and AR (– 59.8 %). ROM of adjacent segments after instrumentation was only
affected in LB, showing a decrease in ROM (−12.9 %). Spinal stiffness of adjacent segments was not affected after instrumentation. In conclusion, while posterior instrumentation, as might be expected, substantially restricts spinal motion the treated segments, it does not result in increased range of motion or decreased stiffness of adjacent segments. Thus, spinal segments proximal to posterior fusion techniques do not seem to progress into unstable segments from an in vitro biomechanical point of view.

Clinical implications for laminectomy surgery

Our data implies that lumbar laminectomy increases the risk of developing a post-laminectomy syndrome during application of high loads, which could be present when performing demanding activities. However, this risk might be assessed pre-operatively since we showed that multiple independent variables, together, largely determine the stiffness and failure strength of a lumbar spinal segment treated with a laminectomy. For both shear and torsion biomechanics, it was found that low vertebral bone content (BMC and BMD), small intervertebral discs, small bony structures, facet joint tropism and absence of osteophytes could predict the possible development of postoperative instability following lumbar laminectomy. For torsion, the decreasing effect of bone quality on spinal strength was even larger than the effects of laminectomy. This information can help to identify which patients are prone to develop a post-laminectomy syndrome and may therefore support or alter surgical decision-making, i.e. the decision whether or not to add additional posterior lumbar instrumentation in addition to a lumbar laminectomy. Some of the measures that we found to predict spinal (failure) biomechanics after lumbar laminectomy can be estimated based on standard pre-operative radiographs, CT- and MRI-imaging. However, standard pre-operative assessment for laminectomy does not include methods that provide the best predictive variables, i.e. DXA measurement to assess bone quality and to quantify bone density. Note that the high predictive value is despite the fact that DXA quantifies overall vertebral bone content rather than specifically for the neural arc. Subjects with low bone quality may require additional posterior instrumented stabilization to prevent postoperative instability. If parameters of bone quality are taken into consideration, often, BMD is used. However, we have also shown that geometry parameters are important for the estimation of post-operative spinal biomechanics. Therefore, especially BMC instead of BMD is a useful parameter since it integrates information on bone density and vertebral dimensions. BMC is defined as BMD (g/cm²) multiplied by the total segmental surface area (FA) of the spinal segment (cm²) and is expressed in grams, leading to the following equation: BMC (g) = BMD (g/cm²) * FA (cm²). Besides the association between bone quality and shear and torsion biomechanics, also compression strength was previously correlated to BMC.
Clinical implications for additional posterior fusion surgery

The risk of developing a post-laminectomy syndrome, including postoperative pars interarticularis fractures, spondylolisthesis and development of degenerative scoliosis, seems quite substantial based on the results of this thesis. It may therefore be questioned whether patients can safely perform physically demanding tasks such as lifting heavy objects from low position after lumbar laminectomy since during these activities high load-levels are found in the lumbar area 13, 14, 15.

Taking this knowledge into consideration, it might be argued that all patients should be provided with posterior instrumentation after performing a laminectomy. However, in addition to well-known complications such as wound infection, the possibility of Adjacent Segment Disease (ASD) is receiving increasing interest among spine surgeons. It is progressively becoming clear that, in particular the use of rigid posterior lumbar interbody fusion (PLIF) and posterior lumbar fusion (PLF) as an addition to lumbar laminectomy holds a risk of ASD 28. Purely pragmatic, typical topping-off procedures are used in some cases to prevent or slow down adjacent segment disease 20. Topping-off procedures combine rigid fusion with a flexible pedicle screw system to prevent ASD 30. During surgical planning, considerations regarding the use and type of posterior instrumentation would be better substantiated when post-operative biomechanical behaviour after laminectomy could be predicted. Currently, a decision whether or not to use additional instrumentation is based upon personal experience. Scientific sound criteria are needed. The current thesis provides biomechanical data that can be used to support such decision-making, as will be outlined in more detail below (Figure 1).

Recommendations and criteria for individual patients

Considering the pros and cons of posterior fusion and in order to provide a recommendation for the use of BMC in clinical practice, a decision making tree for the use of additional instrumentation after a single level lumbar laminectomy for degenerative spinal stenosis, in an individual patient is proposed in Figure 1.

One could argue that a safe strategy would be to focus at a predicted shear and rotational strength after laminectomy that is not lower than the lowest values in the untreated population. We found that shear force to failure (SFF) of untreated segments (5x L2 – L3 and 5x L4 – L5) ranged from 3284 – 909 N. After laminectomy SFF ranged from 1886 – 561 N. Out of ten segments, four segments did not reach a SFF after laminectomy larger than 909 N. Torsion moment to failure (TMF) of untreated segments (5x L2 – L3 and 5x L4 – L5) ranged from 79.2 – 27.8 Nm. After laminectomy TMF ranged from 72.2 – 23.7 N. Out of ten segments, three segments did not reach a TMF after laminectomy larger than 27.8 Nm. Considering corresponding BMC values, the above mentioned criterion would suggest spinal
segments with a BMC < 20 grams should be instrumented. However, note that these data also imply that most segments would not require additional instrumentation after single level lumbar laminectomy.

Although we explanted lumbar spines before DXA measurements, leading to an underestimation of BMD and therefore BMC \(^{31}\), these differences were only small. A definitive BMC cut-off point can be determined more accurately by conducting clinical studies in the future. We recommend assessing spinal BMC to predict clinically relevant instability in a prospective in vivo design to further enhance clinical applicability. In theory, a comparison between in vivo loading and predicted spinal strength could also lead to a clear criterion to decide whether or not a spinal segment progresses into an unstable segment and whether or not spinal instrumentation should be used. However, for this approach there are several issues that will need to be taken into consideration.

First, in vivo loading strongly depends on body weight, length and daily activities, which are all substantially affecting load levels. For shear biomechanics, estimations are inaccurate because it strongly depends on lumbar lordosis. Furthermore, estimations for shear loads strongly depend on segmental level. Shear loads are substantially higher at L5 – S1 than on more proximal situated levels. Therefore, to decide on the need for additional instrumentation the specific segment level should be taken into consideration. For rotation it is unknown what the load levels on segments is in an in vivo situation. Therefore, in addition to clinical studies, also in vivo work to assess load levels per person by adjusting for body-weight and length might strengthen a clinical applicable cut-off value for BMC and should therefore be further investigated.

Second, the use of a BMC-value to improve clinical practice means that all patients undergoing lumbar laminectomy should obtain a DXA-scan before surgery. This low-radiation scan will not only provide a surgeon with the necessary information on post-operative spinal biomechanics, it will also enable a surgeon to foresee possible complications with screw placement and in the long-term pull–out strength of pedicle screws. Pull-out strength of spinal implants is proved to be dependent on bone mineral quality as measured by dual X-ray absorptiometry (DXA) \(^{34}\) and this dependency needs to be taken into account. Techniques to increase pull-out strength in osteoporotic patients such as screw augmentation are available and might be a solution for this specific problem \(^{6}\).

Third, Macintosh and Bogduk elucidated on the biomechanical relevance of the lumbar multifidus muscles and attachments of the lumbar spinous erector muscle, including the longissimus and iliocostalis muscles \(^{21,22}\). These local spinal muscles are crucial for stabilization
of the spine. Basically, the lumbar spine is an instable construct, however, the co-activation of agonist and antagonist muscles increases spinal stability. Due to a laminectomy, all spinal structures attached to the spinal process and part of the lamina are partially disrupted or completely disabled. The addition of spinal instrumentation possibly disrupts the muscular integrity even more. Unfortunately, we were not able to quantify this resultant instability following laminectomy. However, we do suggest that these alterations in \textit{in vivo} muscle function might be crucial for instability and subsequent complications in the long-term, and should thus be investigated in future work.

In Figure 1, a suggested decision making algorithm combining anthropometry, behavioural demands and BMC as an aid to determine spinal stability after a single level lumbar laminectomy is presented. This algorithm can be used as a surgical decision making tool to determine whether or not to use additional instrumentation following single level lumbar laminectomy. At first, a patient is conservatively treated for degenerative lumbar spinal stenosis with the exception of patients with a cauda equina syndrome. In case conservative treatment fails, surgery such as a single level lumbar laminectomy is considered. During surgical planning, patients should have a DXA scan to determine the BMC of the affected spinal segment. In addition, patients should be assessed for criterion adaptation factor A (anthropometry) and factor B (behavioural demands). Since in vivo loading strongly depends on a patient's anthropometry, it might be argued that the initial BMC criterion should be multiplied by factor A. Factor A or body weight and height relative to the population average, could lead to three types of factors including an outcome < 1.0 or 1.0 or > 1.0. Depending on the discrepancy with the population average, the relative factor in- or decreases proportionally. Note that specifically of both body weight and body height increase load levels on the lumbar spine. Therefore, the Body Mass Index (BMI) would not be a suitable parameter. For behavioural demands, a categorical scale could be used. High demanding patients should be ascribed with a factor B > 1.0 while, normal demanding patients would be granted with a factor B = 1.0 and low demanding patients with a factor B < 1.0. Similar to factor A, proportionality of behavioural demands (Factor B) needs to be taken into consideration. Finally, according to the final BMC criterion a more informed choice could be made on the use of instrumentation. Note that this decision-making tree holds a preliminary recommendation. Further research should be conducted in order to, for example, define the effect of body weight and physical activity on in vivo loading of the human lumbar spine.

If in an individual case the use of spinal instrumentation is considered, its primary focus should be on re-stabilizing the level at which the laminectomy was performed, since we did not find substantial effects of single level lumbar laminectomy on adjacent segments in our submaximal test setup. We believe that, for deciding whether or not to add (multilevel) instrumentation
after laminectomy, considerations with regard to failure loads are most important. Our reported effects of laminectomy on failure shear and torsion loads were substantially larger than effects of laminectomy on submaximal biomechanical parameters. Thus, considerations with regard to the question whether or not to apply posterior instrumentation, should be based on failure loads rather than on changes in range of motion and stiffness around the neutral orientation. However, in this thesis we have restricted our results to shear and torsion failure properties. While other motion direction might hold clinical relevance, we believe that shear and torsion loads are in general the most important loading mechanisms leading to spinal trauma.

We also found that posterior instrumentation does not change adjacent level biomechanics with exception of ROM in LB of the proximal levels. Based on our biomechanical results there is no reason for long constructs in single level lumbar laminectomy or the use of topping-off procedures proximally adjacent to rigid fusion. In light of topping-off procedures in addition to posterior fusion, we even advise to be cautious since preventive effects of these procedures on adjacent segment disease have not conclusively been proven. Hence, those procedures do increase the risk of implant related complications. Furthermore, the effects of topping-off procedures on the adjacent levels are unknown. When arguing for the use of topping-off procedures to slow down adjacent segment disease, there does not seem to be a biomechanical argument.

Finally, topping–off procedures perfectly illustrate current changes in spine surgery practice and in healthcare in general. In light of evidence based medicine (EBM) and health technology assessment (HTA), costs of healthcare interventions are increasingly considered important. To be able to make well-informed decisions, it is of utmost importance to obtain valid and reliable information about safety, effectiveness and cost-effectiveness of interventions. It is well known that the use of instrumentation alongside a laminectomy significantly increases the costs of this procedure. In general it could be argued that the clinical advantages of an instrumented procedure should, considering the costs, be substantially better than non-instrumented laminectomy in order to be cost-effective.
**Limitations**

From a biomechanical point of view, this thesis holds several limitations. We acknowledge that our results are mere in vitro biomechanical results and therefore only represent clinical value to certain extent. It is important to emphasize that translation from biomechanical studies to clinical practice is often difficult. However, in vitro evaluation of spinal biomechanics has been acknowledged as the golden standard for many years, especially when it comes to evaluation of surgical strategies and spinal implant development.
When comparing different biomechanical evaluations of the lumbar spine, uniformity of models is crucial in order to come to generalizable conclusions. However, often studies that investigated decompressive techniques and/or instrumentation of the lumbar spine reported substantially different outcomes. In most cases, differences in outcomes were probably caused by more extensive decompressive techniques or the use of follower loads. Another possible explanation was that different load-levels were used. In this thesis, when loading single spine segments, we selected a compression load of 1600 N during failure–testing (Chapters 2 – 5) because it is physiologically relevant 13, 14, 15 and allows for comparison with previous work 33, 35. While this force may seem high, it is not very high compared to that estimated in vivo compression. Mainly due to muscle forces, the spine is already subjected to forces of this magnitude when the trunk is inclined about 45 degrees forward. In addition, this amount of preload is a compromise between applying compression forces that are sufficiently large to simulate spinal loads that occur in vivo when large shear forces are present 13, 14, 15, 32, but low enough to avoid damage due to compression forces alone 1. During our submaximal experiments with complete lumbar spines (Chapters 6 – 9) we preloaded the spine with 250 N for 1 hour, which is obviously relatively low in contrast to physiological loading 1. Preloading with 250 N was chosen, since buckling might occur when loading a complete lumbar spine with a higher load 25. We did not apply axial loading during testing, again to prevent buckling. While so-called follower loads would allow for axial loading during bending without buckling, such loads inevitably cause additional moments of unknown magnitude, which would interfere with the purpose of this thesis. Since we tested spines of elderly cadavers, which could be osteoporotic, we decided to use a 4 Nm load level. It was recommended to decrease standard pure moments of 7.5 Nm by 50 % when testing osteoporotic spines 37. Comparable load levels were previously used 3, 4, 10.

In this thesis we defined spinal instability from a biomechanical perspective as a significant and substantial increase in range of motion (ROM), and/or reduced stiffness (around the neutral zone 23, 24 in submaximal testing experiments) and/or a decrease in ultimate spinal strength. In addition, we have added yielding or a derivative of yielding (i.e. early torsion stiffness versus late torsion stiffness) to these commonly used biomechanical parameters in our biomechanical evaluations (Chapters 3 – 5). The yield point might be one of the most critical biomechanical properties, because it marks the beginning of the irreversible deformation of a spinal motion segment, signaling the appearance of the soft tissue and / or trabecular bone lesions 29. We expect that when load levels cross the yield point, sub-clinical damage will occur. Such damage may, at a later stage, lead to symptoms. Unfortunately, we were not able to exactly calculate the yield point in torsion testing. Therefore we argued that the yield point was within the transition zone between ETS and LTS. Obviously, a yield point was not defined during submaximal testing experiments.
Another limitation regarding the use of posterior instrumentation is that we did not investigate PLIF, TLIF and XLIF techniques. Replacement of the intervertebral disc by interbody devices, such as a cage, might lead to different biomechanical outcomes. We cannot draw conclusions on this matter and recommend a new biomechanical study to evaluate these effects.

Conclusions
In conclusion, single level facet sparing lumbar laminectomy destabilizes the spine. To put these biomechanical alterations in clinical perspective is a challenge. However, we found that a pre-operative DXA scan to quantify BMC provides a surgeon with a practical tool to prognosticate which spinal segments have a high risk of developing into an unstable one after performing a single level lumbar laminectomy. In most cases instrumentation as an additive to single level lumbar laminectomy will prove not be necessary because residual spinal stability will fall within our proposed criteria. In case it is decided to use additional instrumentation then our study results show that it is sufficient to stabilize the decompressed segment only. A laminectomy does not seem to alter the biomechanical behavior of the adjacent segment, however the possible role of post-operative muscle function should thereby be considered and requires further evaluation in future research. Furthermore, in a situation in which a single level lumbar laminectomy is instrumented, there does not seem to be a substantial alteration in the biomechanical behavior of the adjacent level. Based on these results, it could be argued that, in case of a single level decompression with additional instrumentation there is no need to extent the spondylodeses with rigid or topping-off dynamic instrumentation techniques.
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General discussion | 161
Nederlandse samenvatting
Relevante van het proefschrift

In de orthopedische en neurochirurgische praktijk worden vaak oudere patiënten gezien met klachten behorende bij symptomatische degeneratieve lumbale spinale stenose. In 1950 was Verbiest de eerste die beschreef dat de neurologische symptomen een gevolg van lumbale stenose en daarmee van neurale compressie waren. De specifieke symptomen betreffen (intermitterende) pijn in de benen, tintelingen, gevoelloosheid en spierzwakte. Klachten nemen vaak af wanneer een patiënt voorover buigt. Voorafgaand hebben patiënten veelal een lange historie van lage rugklachten. Er zijn meerdere pathofysiologische mechanismen die kunnen leiden tot deze symptomen. Degeneratieve afwijkingen, spondylolisthesis en (osteoporotische) fracturen kunnen allemaal compressie op neurologische structuren veroorzaken. In dit proefschrift ligt de nadruk op de chirurgische behandeling van symptomatische degeneratieve lumbale spinale stenose door middel van een laminectomie, ongeacht de pathogenese. Na het stellen van de diagnose symptomatische lumbale degeneratieve spinale stenose heeft een patiënt in het algemeen drie behandelingen, te weten niet-operatieve behandelingen, chirurgische decompressie (meestal een laminectomie) en spinale decompressie chirurgie in combinatie met instrumentatie.

Wanneer een patiënt neurologische klachten heeft bij een degeneratieve lumbale spinale stenose dan wordt hij of zij eerst conservatief behandeld. Conservatieve behandeling kan bestaan uit één of meer van de volgende componenten: het vermijden van pijnlijke activiteiten, educatie, pijnstilling en bewegings- of fysiotherapie Indien er zogenaamde alarmsymptomen (bijvoorbeeld het cauda equina syndroom) aanwezig zijn wordt er uiteraard chirurgisch ingegrepen.

Wanneer conservatieve behandeling faalt wordt in de meeste gevallen een chirurgische ingreep overwogen. Bij patiënten ouder dan 65 jaar is symptomatische degeneratieve lumbale spinale stenose de meest voorkomende indicatie voor spinale chirurgie. Diverse decompressietechnieken, zoals laminectomie, laminotomie, foraminotomie en facetectomy zijn in het verleden bewezen effectief bevonden. Deze technieken worden allen gebruikt om gecomprimeerde zenuwwortels of de cauda equina te decomprimeren en daarmee de symptomen te verlichten. Onder deze vormen van decompressie is facet sparinge laminectomie de meest gebruikte chirurgische interventie voor degeneratieve lumbale spinale stenose. Na laminectomie verdwijnen over het algemeen de (intermitterende) neurologische symptomen, zoals pijn in de benen, tintelingen, gevoelloosheid en zwakte in grotere mate dan wanneer conservatieve behandeling wordt voortgezet. Bovendien is aangetoond dat de kwaliteit van leven na een decompressie laminectomie toeneemt.
Hoewel een laminectomie in de meeste gevallen de symptomen verlicht, wordt door deze techniek de anatomie veranderd, waardoor ook de biomechanica van de lumbale wervelkolom beïnvloedt wordt. Veranderende biomechanica van de wervelkolom na laminectomie kan leiden tot symptomatiche postoperatieve lumbale instabiliteit of postoperatief falen van het behandelde spinale segment, ook wel het post-laminectomie syndroom genoemd. De meest frequent gerapporteerde complicaties na laminectomie die in de literatuur worden gemeld zijn postoperatieve segmentale instabiliteit van de lumbale wervelkolom of postoperatieve iatrogene spondylolisthesis. Symptomatische spondylolisthesis rechtvaardigt een heroperatie waarbij het instabiele segment wordt gestabiliseerd met als uiteindelijk doel fusie van het segment. De incidentie van spondylolisthesis of het zogenaamde post-laminectomie syndroom na een facet sparende laminectomie varieert van 8 % - 31 %.

Hoewel de facetgewrichten vaak behouden blijven bij een laminectomie worden de lamina, interspinale-, supraspinale- en flavum ligamenten ingesneden en/of verwijderd. Gezien het feit dat deze structuren een stabiliserende functie hebben en ondersteuning bieden aan diverse spiergroepen kan verwijdering of beschadiging van deze structuren tot klinisch relevante spinale instabiliteit, vervorming (bijvoorbeeld scoliose of spondylolisthesis) of pijn leiden. De biomechanische effecten van een laminectomie op één niveau zijn echter nog niet grondig onderzocht. Daarnaast is het niet bekend of deze effecten kunnen worden voorspeld door preoperatieve identificatie van segmenten die een hoge kans hebben op progressie tot een klinische instabiliteit. Als dit mogelijk zou zijn zouden chirurgen voorafgaand aan de operatie kunnen voorspellen welke patiënten baat kunnen hebben bij het tijdens de laminectomie operatie plaatsen van aanvullende instrumentatie om het gedecomprimeerde segment te stabiliseren.

Momenteel zijn er geen strikte richtlijnen met betrekking tot de vraag wanneer aanvullende instrumentatie dient te worden gebruikt. In de operatiekamer is de ervaring van de chirurg tot op heden beslissend. In het geval dat er instrumentatie wordt gebruikt betreft het meestal een posterieure pedikelschroef/staaf-constructie met daarbij een botplastiek. Posterieure fusie wordt beschouwd als standaard chirurgisch handeling die kan worden toegepast bij een gedecomprimeerde lumbale wervelkolom waarbij het vermoeden bestaat dat zich postoperatieve klinisch instabiliteit kan ontwikkelen. Bij het gebruik van posterieure instrumentatie streeft een chirurg naar een arthrodese waarbij fusie van een segment plaatsvindt door botvorming. Succesvolle arthrodese van een segment na posterieure lumbale fusie wordt gezien bij ongeveer 90 % van de patiënten en leidt tot goede klinische resultaten.
Echter, het gebruik van instrumentatie verhoogt de kans op implantaat gerelateerde complicaties, zoals infecties, neurologische stoornissen, verhoogd bloedverlies tijdens de operatie, langere duur van de ingreep, breken of loslating van instrumentatie en op problemen in aangrenzende segmenten, oftewel _adjacent segment disease_ (ASD) genoemd. De kans op implantaat gerelateerde complicaties moet daarom worden afgewogen tegen het risico van postoperatieve complicaties na laminectomie zonder stabilisatie. Tenslotte moet worden opgemerkt dat het gebruik van instrumentatie de kosten van de operatie ook aanzienlijk verhoogt.

Gezien het bovenstaande is het van belang dat de chirurg beter in staat wordt gesteld om een wetenschappelijk onderbouwde keuze te maken tussen het verrichten van een ongeïnstrumenteerde en geïnstrumenteerde laminectomie bij de individuele patiënt. Daarnaast is het van belang dat hij of zij inzicht krijgt in de effecten van instrumentatie op de lumbale wervelkolom.

_Doelstellingen proefschrift_
In de hoofdstukken 2 tot en met 8 worden in totaal zeven experimentele studies gepresenteerd. De hoofdstukken 2 tot en met 5 beschrijven experimenten tot falen op een enkel niveau (L2 - L3 of L4 - L5), terwijl in de hoofdstukken 6 tot en met 8 submaximale testen van de gehele lumbale wervelkolom (L1 - L5) worden beschreven. Bij alle experimentele studies werd gebruik gemaakt van vers ingevroren humane lumbale wervelkolommen.

De doelen van dit proefschrift zijn om:
1. De effecten van een laminectomie op de biomechanica bij afschuifkrachten op het behandelde segment te onderzoeken (Hoofdstuk 2).
2. Te onderzoeken of het biomechanisch gedrag bij afschuifkrachten na laminectomie kan worden voorspeld (Hoofdstuk 3).
3. De effecten van een laminectomie op torsie biomechanica van het behandelde segment (Hoofdstuk 4) te onderzoeken.
4. Te onderzoeken of torsie biomechanica na laminectomie kan worden voorspeld (Hoofdstuk 5).
5. De methodologie voor submaximale biomechanische testen van spinale segmenten (Hoofdstuk 6) te verbeteren.
6. De effecten van laminectomie op het submaximale biomechanisch gedrag van behandelde en aangrenzende segmenten (Hoofdstuk 7) te onderzoeken.
7. De effecten van geïnstrumenteerde laminectomie op het submaximale biomechanisch gedrag van behandelde en aangrenzende segmenten (Hoofdstuk 8) te onderzoeken.
Op basis van de resultaten van de voorgaande hoofdstukken wordt in de discussie (Hoofdstuk 9) een stroomdiagram gepresenteerd dat kan helpen bij de chirurgische besluitvorming ten behoeve van het handelen bij een decompressie laminectomie.

Samenvatting van hoofdstukken

In het 2e en 3e hoofdstuk bleek dat laminectomie resulteert in een afname van afschuifsterkte (44.2 %), afname van de afschuiving waarbij het segment faalt (38.6 %), en afname van de afschuivingsstijfheid (19.9 %). Verder nam de yield force, de kracht waarbij overgang van elastische tot plastische vervorming plaats vindt, met 41.1 % af na laminectomie. Voor segmenten waarop een laminectomie werd uitgevoerd, was de afschuivingsstijfheid significant gecorreleerd met de mate van degeneratie van de tussenwervelschijf en van de facetgewrichten. De yield force was gecorreleerd met tussenwervelschijf geometrie (lengte, oppervlakte en volume), botmineraalgehalte (BMC) en frontaal oppervlak. De sterkte van het spinale segment bij falen bleek gecorreleerd te zijn met de voor-achterwaartse lengte van de tussenwervelschijf, BMC en de botmineraaldichtheid (BMD). Met de bovengenoemde variabelen konden de afschuivingsstijfheid, de yield force en de kracht tot falen na laminectomie goed worden voorspeld (r²-waarden waren respectievelijk 0.53, 0.81 en 0.77). Er werd geconcludeerd dat verlies van afschuif sterkte en stijfheid redelijk voorspeld kan worden door combinatie van DXA-afgeleide maten van botkwantiteit en -kwaliteit (respectievelijk BMC en BMD), van geometrie van de tussenwervelschijf en van degeneratieve parameters. Dit suggereert dat een lage BMC of BMD, kleine tussenwervelschijven en osteofyten de mogelijke ontwikkeling van postoperatieve afschuifinstabiliteit na lumbale laminectomie kunnen voorspellen.

In de hoofdstukken 4 en 5 bleek dat bij torsie-belasting de belasting-verplaatsingscurves een typische bifasisch patroon hebben met een duidelijk verschil tussen torsiestijfheid in de beginfase en in de late fase. Na laminectomie werd een aanzienlijke daling van de stijfheid gevonden, zowel in de beginfase (34.1 %) als in de late fase (30.1 %). Het torsiemoment tot falen daalde in mindere mate (17.6 %) na laminectomie. Maten voor botkwaliteit, een reeks van geometrische kenmerken, en een aantal maten van degeneratie van de tussenwervelschijf bleken geassocieerd te zijn met de biomechanische eigenschappen van de lumbale segmenten in torsie. Multivariate analyses toonden aan dat met een combinatie van bovengenoemde maten de beginfase en late fase stijfheid en het torsiemoment tot falen zeer goed waren te voorspellen voor segmenten die werden behandeld met een laminectomie (r²-waarden waren respectievelijk 0.95, 0.87 en 0.93). Hierbij waren, net als bij afschuiving, BMC en BMD de belangrijkste voorspellers. Bovendien bleken de geometrie van de tussenwervelschijf en het verschil in de hoek van de linker en rechter facetgewrichten ook voorspellend te zijn. Er werd geconcludeerd dat, net als voor afschuifbiomechanica, DXA-afgeleide bot parameters
Hoofdstuk 6 beschrijft een methodologische studie naar het effect van herhaling van belastingscycli op biomechanische parameters. De studie is bedoeld als aanvulling op de huidige gouden standaard, die wordt gebruikt bij submaximale testprocedures voor de lumbale wervelkolom in flexie en extensie (FE), laterale buigingen (LB) en axiale rotatie (AR). Gewonden werd dat bij de opgelegde belasting het bewegingsbereik de range of motion (ROM) absoluut gezien in kleine, maar significant mate toenam in alle richtingen na drie (FE: 1.0 %, LB: 1.5 % en AR: 1.5 %) en na tien testcylci (FE: 2.9 %, LB: 3.3 % en AR: 3.3 %). Stijfheid, gemeten rond de neutrale zone, werd niet significant beïnvloed maar varieerde aanzienlijk gedurende de testcylci. Er werd geconcludeerd dat, hoewel de effecten klein waren, de beoordeling van de tiende cyclus in plaats van de gebruikelijke derde cyclus de mogelijke visco-elastische effecten bij herhaalde metingen van de ROM vermindert, omdat de wervelkolom zich dichterbij een evenwichtssituatie bevindt. Bovendien zou de beoordeling van stijfheid kunnen worden verbeterd door een gemiddelde van tien testcylci te nemen.

Hoofdstuk 7 liet zien dat bij het verrichten van submaximale testen, de ROM op het niveau van laminectomie significant toenam in FE (7.3 %), LB (7.5 %) en AR (12.2 %). De ROM van aangrenzende segmenten werd niet significant beïnvloed door een laminectomie, met uitzondering van LB (-7.7 %). Controle segmenten werden niet beïnvloed. De stijfheid rond de neutrale zone van de behandelde, aangrenzende en controle segmenten werd niet beïnvloed door laminectomie.

In hoofdstuk 8 werden soortgelijke testen uitgevoerd als in hoofdstuk 7, maar nu werden de testen ook herhaald na toepassing van instrumentatie. Hoewel de resultaten enigszins afwijken van hoofdstuk 7, bleek, net als in hoofdstuk 7, dat een laminectomie slechts beperkte gevolgen had voor de submaximale biomechanica van het behandelde segment en had deze ingreep een nog kleiner effect op aangrenzende segmenten. In de behandelde segmenten nam de ROM in AR significant toe (19.4 %) en de stijfheid rond de neutrale zone nam significant af (-18.0 %) na laminectomie, terwijl er geen significant effect na laminectomie op FE en LB werd gezien. In de aangrenzende segmenten nam de ROM in AR enigszins toe (11.0 %) terwijl de ROM in FE en LB niet veranderde. Na het aanbrengen van posterieure instrumentatie verminderde de ROM sterk in FE (-74.3 %), LB (-71.6 %) en AR (-59.8 %). De ROM van aangrenzende segmenten na instrumentatie werd alleen beïnvloed in LB waarbij er een afname in ROM werd gezien van 12.9 %. De spinale stijfheid van aangrenzende segmenten werd niet beïnvloed door instrumentatie. Geconcludeerd werd dat, terwijl enerzijds posterieure instrumentatie, zoals verwacht, de beweeglijkheid van de behandelde
segmenten sterk reduceerde, dit niet leidde tot veranderingen in de beweeglijkheid of stijfheid van aanliggende segmenten. Er blijkt dan ook geen biomechanische argumentatie te bestaan om proximaal of distaal aan een posterieure fusie additionele stabilisatie toe te passen om het ontstaan van instabiliteit te voorkomen.

**Interpretatie**

In de orthopedische en neurochirurgische praktijk worden oudere patiënten met symptomatiche degeneratieve lumbale spinale stenose vaak behandeld met een laminectomie. Postoperatieve symptomatische klinische instabiliteit die na deze ingreep kan ontstaan rechtvaardigt re-operatie om het instabiele segment te stabiliseren en te fuseren. Hoewel spinale stabiliteit toeneemt na het plaatsen van posterieure instrumentatie, veroorzaakt de procedure van stabilisatie zelf, volgens de klinische literatuur een verhoogde kans op implantaat gerelateerde complicaties, waaronder ASD. Bovendien verhoogt het gebruik van posterieure instrumentatie de kosten van de operatie aanzienlijk.

Dit proefschrift is het eerste complete overzicht van spinale biomechanica na een lumbale (geinstrumenteerde) laminectomie op één niveau. In het geval van deze specifieke decompressie techniek werden in de literatuur over het algemeen experimenten op diermodellen of eindige elementen modellen beschreven. Bovendien werden er in het verleden in hoofdzaak resultaten van uitgebreide decompressie technieken gepubliceerd en werd instabiliteit van het aangrenzende segment na laminectomie slechts in een zeer beperkt aantal studies beschreven. Literatuur met betrekking tot de biomechanische effecten van aangrenzende segmenten na rigide posterieure instrumentatie is eveneens schaars.

Op basis van de gepresenteerde resultaten in dit proefschrift kan gesteld worden dat er biomechanische redenen zijn waarom, na laminectomie, de kans op het ontwikkelen van een post-laminectomie syndroom aanzienlijk is. Men kan zich daarom afvragen of patiënten, na lumbale laminectomie, veilig fysiek veeleisende taken, zoals het tillen van zware voorwerpen vanaf de grond, kunnen uitvoeren. Tijdens deze activiteiten ontstaat er immers een hoge belasting in het lumbale gebied. Aan de andere kant staat de mogelijkheid van het ontstaan van ASD, na het verrichten van een posterieure fusie, steeds vaker in de belangstelling bij wervelkolomchirurgen.

In de operatiekamer zou een chirurg keuzes met betrekking tot de eventuele toepassing en het specifieke type posterieure instrumentatie beter kunnen onderbouwen wanneer hij of zij postoperatief biomechanisch gedrag na laminectomie zou kunnen voorspellen. De patiënt zou hiervan kunnen profiteren. Momenteel beslissen wervelkolomchirurgen om al dan niet om extra instrumentatie technieken bij een laminectomie te gebruiken op basis van persoonlijke ervaring.
De data die in dit proefschrift worden gepresenteerd, suggereren dat laminectomie een patiënt in een risicovolle situatie brengt ten aanzien van het ontwikkelen van een post-laminectomie syndroom, met name bij het uitvoeren van activiteiten die een zware belasting van de rug opleveren. Naast de biomechanische factoren, die in dit proefschrift worden behandeld, is er nog een reden dat laminectomie de kans op een post-laminectomie syndroom verhoogt: bij het verrichten van een laminectomie worden lokale spinale spieren (deels) verrijkerd terwijl deze spieren essentieel zijn voor het stabiliseren van de wervelkolom. De toevoeging van spinale instrumentatie verstoort de integriteit van de spieren mogelijk nog meer. Veranderingen in de in vivo spierfunctie zijn mogelijk ook mede een oorzaak voor postoperatieve instabiliteit en daarmee voor daaropvolgende complicaties op de lange termijn.

De gegevens uit dit proefschrift kunnen gebruikt worden om het biomechanische risico op een post-laminectomie syndroom preoperatief mede te schatten, omdat er is gebleken dat meerdere onafhankelijke variabelen samen in hoge mate de stijfheid en de faalsterkte van lumbale spinale segment na een laminectomie bepalen. Voor zowel biomechanisch gedrag in de afschuifrichting als voor torsie-biomechanica bleek dat segmenten met een lage botkwaliteit, kleine tussenwervelschijven, kleine afmetingen van benige structuren, verschil tussen linker en rechter hoeken van de facetgewrichten en osteofyten de stijfheid en sterkte na lumbale laminectomie kunnen voorspellen. Voor torsie was het effect van botkwaliteit op spinale sterkte zelfs groter dan de effecten van laminectomie. Deze informatie kan de chirurg helpen om te bepalen welke patiënten een grote kans maken op het ontwikkelen van een post-laminectomie syndroom, waardoor de keuze van de chirurgische techniek reeds tijdens de preoperatieve planning kan worden bepaald.

Enkele van de maten die in dit proefschrift werden gebruikt om de spinale biomechanica na een lumbale laminectomie te voorspellen, kunnen worden bepaald op basis van de standaard preoperatieve röntgenfoto’s, CT- en MRI-scans. Echter, de standaard preoperatieve beeldvorming voor laminectomie bevat niet de methode die de best voorspellende variabelen oplevert, namelijk een DXA-meting om botdichtheid te kwantificeren. Bij patiënten met een matige tot slechte botkwaliteit kan aanvullende posterieure geïnstrumenteerde stabilisatie nodig zijn om postoperatieve instabiliteit te voorkomen. Wanneer, in de klinische praktijk, botkwaliteit-parameters gebruikt worden, betreft het vaak een BMD-waarde. Dit proefschrift heeft echter aangetoond dat de geometrische parameters ook van belang zijn bij de schatting van postoperatieve spinale biomechanica. Om die reden is voor laminectomie de BMC een betere parameter dan BMD om risico’s in te schatten, omdat de BMC zowel vertebrale botdichtheid als afmetingen integreert in één waarde. BMC wordt immers gedefinieerd als BMD (g/cm²) vermenigvuldigd met het totale segmentale oppervlak van de spinale segment
Naast de associatie tussen botkwaliteit en afschuiving- en torsie biomechanica werd in eerdere studies ook axiale compressiesterkte van spinale segmenten gecorreleerd met BMC.

Om een richtlijn voor het gebruik van BMC in de klinische praktijk te bieden, zou men kunnen stellen dat het een veilige strategie zou zijn om uit te gaan van een voorspelde afschuivings- en rotatiesterkte na laminectomie die tenminste even groot is als de laagste waarde in de onbehandelde populatie. We vonden dat schuifkracht tot falen van onbehandelde segmenten varieerde van 3284 - 909N. Na laminectomie varieerde dit van 1886 - 561 N bij een totaal van tien segmenten. Vier segmenten hadden na laminectomie een sterke die kleiner was dan 909 N. De torsie sterkte van onbehandelde segmenten varieerde 79.2 – 27.8 Nm. Na laminectomie varieerde deze sterkte van 72.2 – 23.7 Nm bij een totaal van tien segmenten. Drie segmenten hadden na laminectomie een sterke kleiner dan 27.8 Nm. Gezien de overeenkomstige BMC-waarden bij de zwakste segmenten in beide experimenten zou de drempel voor instrumentatie, gebaseerd op het voorgestelde criterium, op een BMC < 20 g/cm² komen te liggen. Een definitief BMC afkappunt kan nauwkeuriger worden bepaald door het uitvoeren van klinische studies in de toekomst. Daarnaast zijn er verschillende zaken, die in acht moeten worden genomen, wanneer men een dergelijk afkappunt introduceert. Immers, het risico op falen hangt niet alleen af van de belastbaarheid maar ook van de belasting. De belasting van de rug in het dagelijks leven zal sterk afhangen van het lichaamsgewicht, de lichaamslengte en dagelijkse activiteiten.

Praktisch gezien betekent het gebruik van een BMC-waarde in de klinische praktijk dat alle patiënten waarbij een lumbale laminectomie wordt verricht, preoperatief een DXA-scan moeten ondergaan. Deze scan zal een chirurg niet alleen de nodige informatie over postoperatieve spinale biomechanica verschaffen, maar hem of haar ook in staat stellen om mogelijke complicaties met schroef plaatsing en op de lange termijn pull-out kracht van pedikelschroeven te voorzien. De zogenaamde pull-out kracht van spinale implantaten bleek eerder sterk afhankelijk te zijn van de BMD.

Indien bij een patiënt het gebruik van spinale instrumentatie noodzakelijk wordt geacht, dan zou de primaire focus moeten zijn re-stabilisatie van het niveau waarop de laminectomie werd uitgevoerd. Immers, er werden geen substantiële effecten van een lumbale laminectomie op aangrenzende segmenten gevonden. Overwegingen met betrekking tot postoperatief falen van een segment zijn, gezien de resultaten van dit proefschrift, hierbij het meest belangrijk. Immers, de in dit proefschrift gerapporteerde effecten van laminectomie op maximale afschuif- en torsie belastbaarheid waren aanzienlijk groter dan de effecten van laminectomie op submaximale biomechanische parameters. Gezien de afwezigheid van effecten op aangrenzende segmenten werd er, vanuit biomechanisch oogpunt, geen reden gevonden
voor het gebruik van lange constructen bij een lumbale laminectomie op één niveau of voor het gebruik van *topping-off* procedures proximaal bij een rigide posterieure fusie. *Topping-off* procedures combineren rigide fusie met een flexibel pedikelschroefsysteem. Ook werden in de literatuur geen aanwijzingen gevonden dat *topping-off* procedures naast posterieure fusie een preventief effect hebben op ASD. Dergelijke procedures verhogen het risico van implantaat gerelateerde complicaties echter wel. Er lijkt op basis van dit proefschrift geen biomechanische argument te bestaan die pleit voor een oorzakelijk verband tussen rigide fusie bij laminectomie en het bestaan van ASD.

Tot slot, de discussie rond *topping-off* procedures illustreert de huidige veranderingen binnen de wervelkolomchirurgie en in de gezondheidszorg. In het licht van *Evidence Based Medicine* en *Health Technology Assessment*, worden de kosten van de gezondheidszorg interventies in toenemende mate van belang geacht. Gezien het feit dat een chirurg in staat moet zijn om goed onderbouwde beslissingen te nemen, is het van het grootste belang om valide en betrouwbare informatie over veiligheid, effectiviteit en kosteneffectiviteit van interventies te verkrijgen. Het is bekend dat het gebruik van instrumenten naast een laminectomie de kosten van de procedure verhoogt. In het algemeen moet worden gesteld dat de klinische effecten van een geïnstrumenteerde procedure, rekening houdend met de kosten, aanzienlijk beter moeten zijn dan een niet-geïnstrumenteerde laminectomie wil de ingreep aan de richtlijnen van *Evidence Based Medicine* en *Health Technology Assessment* voldoen.

**Conclusies**

De belangrijkste conclusie van dit proefschrift is dat facet sparende lumbale laminectomie op één spinaal niveau de wervelkolom destabiliseert. Om deze biomechanische veranderingen in klinisch perspectief te zetten is een uitdaging. Toch vonden we dat een preoperatieve DXA-scan om de BMC te kwantificeren de chirurg een praktisch instrument kan bieden om te voorspellen welke patiënten een hoog risico hebben op het ontwikkelen van een post-laminectomie syndroom. Er werd hiertoe een in de kliniek toepasbare beslisboom voorgesteld. De beslisboom stelt voor dat de chirurg, bij falend conservatieve behandeling, een DXA scan laat maken en vervolgens de BMC van de patiënt vergelijkt met een drempel. Bij een te lage BMC besluit de chirurg volgens de beslisboom tot aanvullende stabilisatie. Hierbij zou de startwaarde van de drempel 20 gram kunnen zijn, waarbij deze drempel eerst nog wel aangepast wordt aan de feitelijke belasting van de patiënt. Als een chirurg besluit aanvullend stabiliserende technieken te gebruiken dan lijkt het voldoende om alleen het gedeprimeerde segment te stabiliseren. Een laminectomie lijkt immers niet het biomechanische gedrag van de aangrenzende segmenten te veranderen. Gebaseerd op deze resultaten kan er worden gesteld dat bij geïnstrumenteerde decompressie op één niveau er in principe geen reden is voor een wervelkolomchirurg om de spondylodeses aan te vullen met een rigide of *topping-off* systeem.
Appendices

List of abbreviations

Contributors

Commission

Acknowledgements

List of publications related to this thesis

List of presentations and awards related to this thesis

About the author
### List of abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>ANOVA</td>
<td>Analysis of variance</td>
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<td>AR</td>
<td>Axial rotation</td>
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<td>ASD</td>
<td>Adjacent segment disease</td>
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<tr>
<td>BMC</td>
<td>Bone mineral content</td>
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<tr>
<td>BMD</td>
<td>Bone mineral density</td>
</tr>
<tr>
<td>CV</td>
<td>Continuous variable</td>
</tr>
<tr>
<td>DF</td>
<td>Displacement at failure</td>
</tr>
<tr>
<td>DV</td>
<td>Dichotomized variable</td>
</tr>
<tr>
<td>DXA</td>
<td>Dual X-ray absorptiometry</td>
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<tr>
<td>EBM</td>
<td>Evidence based medicine</td>
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<td>ETS</td>
<td>Early torsion stiffness</td>
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<td>FA</td>
<td>Frontal area</td>
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<td>FE</td>
<td>Flexion and extension</td>
</tr>
<tr>
<td>GEE</td>
<td>Generalized estimating equations</td>
</tr>
<tr>
<td>HTA</td>
<td>Health technology assessment</td>
</tr>
<tr>
<td>LB</td>
<td>Lateral bending</td>
</tr>
<tr>
<td>LTS</td>
<td>Late torsion stiffness</td>
</tr>
<tr>
<td>OV</td>
<td>Ordinal variable</td>
</tr>
<tr>
<td>PLF</td>
<td>Posterior lumbar fusion</td>
</tr>
<tr>
<td>PLIF</td>
<td>Posterior lumbar interbody fusion</td>
</tr>
<tr>
<td>QoL</td>
<td>Quality of life</td>
</tr>
<tr>
<td>ROM</td>
<td>Range of motion</td>
</tr>
<tr>
<td>SD</td>
<td>Standard deviation</td>
</tr>
<tr>
<td>SFF</td>
<td>Shear force to failure</td>
</tr>
<tr>
<td>SS</td>
<td>Shear stiffness</td>
</tr>
<tr>
<td>SYF</td>
<td>Shear yield force</td>
</tr>
<tr>
<td>TMF</td>
<td>Torsion moment to failure</td>
</tr>
<tr>
<td>TS</td>
<td>Torsion stiffness</td>
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Publications related to this thesis

1. **The impact of bone mineral density and disc degeneration on shear strength and stiffness of the lumbar spine following laminectomy.**

2. **Which factors prognosticate spinal instability following lumbar laminectomy?**

3. **Torsion biomechanics of the spine following lumbar laminectomy: a human cadaver study.**

4. **Which factors prognosticate rotational instability following lumbar laminectomy?**

5. **Effects of repetitive movement on range of motion and stiffness around the neutral orientation of the human lumbar spine.**

6. **Single level lumbar laminectomy alters segmental biomechanical behavior without affecting adjacent segments.**

7. **The effects of single level instrumented lumbar laminectomy on adjacent spinal biomechanics.**
   *Arno Bisschop, Roderick M. Holewijn, Ildsart Kingma, Agnita Stadhouder, Pieter-Paul A. Vergroesen, Albert J. van der Veen, Jaap H. van Dieën & Barend J. van Royen.* Accepted on October 8th for publication in Global Spine Journal - in press
Presentations and awards related to this thesis

Contents of this thesis were presented at the following congresses and symposia

BIOSPINE 3 (2010) – podium presentation
Amsterdam, The Netherlands

Kyoto, Japan

Symposium Experimenteel Onderzoek Heelkundige Specialismen (SEOHS) (2010) – poster presentation
Rotterdam, The Netherlands

Honours Symposium (2010) – poster presentation
Amsterdam, The Netherlands

Annual Meeting of the Nederlandse Orthopaedische Vereniging (NOV) (2011) – podium presentation
Groningen, The Netherlands

Studenten onderzoeks prijs (2011) – podium presentation
Amsterdam, The Netherlands

Orthopedic Research Society (ORS) (2012) – poster presentation
San Francisco, United States of America

Orthopedic Research Society (ORS) (2012) – poster presentation
San Francisco, United States of America

International Society for the Advancement of Spine Surgery (ISASS) (2012) – podium presentation
Barcelona, Spain

International Society for the Advancement of Spine Surgery (ISASS) (2012) – podium presentation
Barcelona, Spain
Amsterdam, The Netherlands

Eurospine (2012) – podium presentation
Amsterdam, The Netherlands

Nijmegen, The Netherlands

Global Spine Congress (GSC) (2013) – podium presentation
Hong Kong, Hong Kong SAR PRC

Global Spine Congress (GSC) (2013) – poster presentation
Hong Kong, Hong Kong SAR PRC

Spine and Wine Symposium (2014) – podium presentation
Amsterdam, The Netherlands

Seoul, South Korea

Awards

Runner up Honours Awards (2010)
Runner up Annual Student Research Award (2011)
Annual Dutch Spine Society Award (2012)
Spinal stability following single level lumbar laminectomy:

A biomechanical study of the human spine

About the author

Arno Bisschop was born in Hoorn, The Netherlands on February 12, 1987. He completed his secondary education at the Han Fortmann College (Heerhugowaard) in 2005. During his high-school period, he pursued a sportive career in speed skating and competed in multiple Dutch championships. In September 2005, he started with his bachelor in biomedical sciences at the VU university. After obtaining his propaedeutic examination, he commenced his bachelor in medicine in 2006. Due to his sportive career, he developed a keen interest for orthopedic surgery. As a result, he started his research at the department of orthopedic surgery in the VU university medical center within the Honours program. Later, this research evolved into a PhD position under supervision of prof. dr. B.J. van Royen, prof. dr. J.H. van Dieën and dr. I. Kingma. In the mean time, he completed his minor in entrepreneurship and started working as an anatomy teacher at the department of anatomy at the VU university medical center and as surgical assistant at the department of orthopedic surgery in Het Kennemer Gasthuis in Haarlem.

Besides his orthopedic and sportive ambitions, his interests in healthcare economics and healthcare-based entrepreneurship also grew during this period. As a result, he co-founded the Medical Business foundation and several start-up companies.

Just recently he obtained his master in medicine with honours and his master in management, policy-analysis and entrepreneurship in health and life sciences. During the latter, he performed research on The Spine Clinic, a conceptual model of a privately held clinic focusing on value-based spine care, innovative techniques and cost-effectiveness of treatment strategies. In addition, he wrote a thesis on possible improvements of the FDA regulation for new spinal implants.

In the future, he hopes to combine clinical work with his business and entrepreneurial background. In addition, he will continue his research on spinal biomechanics.
Invitation

For the public defense of the PhD thesis of Arno Bisschop entitled:

Spinal stability following single level lumbar laminectomy: A biomechanical study of the human spine


The reception will be held afterwards in The Basket at the Vrije Universiteit campus.

Questions related to the PhD defense can be addressed to my paranymphs, Kees-Pieter Paul and Puck Vergroesen (arno.promotie@gmail.com).

I hope to welcome you at the ceremony and to meet you at the reception!

Best regards,

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