The effect of design parameters of dynamic pedicle screw systems on kinematics and load bearing: an in vitro study

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Received: 4 May 2010 / Revised: 29 September 2010 / Accepted: 31 October 2010 / Published online: 26 November 2010
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Abstract As an alternative treatment for chronic back pain due to disc degeneration motion preserving techniques such as posterior dynamic stabilization (PDS) has been clinically introduced, with the intention to alter the load transfer and the kinematics at the affected level to delay degeneration. However, up to the present, it remains unclear when a PDS is clinically indicated and how the ideal PDS mechanism should be designed to achieve this goal. Therefore, the objective of this study was to compare different PDS devices against rigid fixation to investigate the biomechanical impact of PDS design on stabilization and load transfer in the treated and adjacent cranial segment. Six human lumbar spine specimens (L3–L5) were tested in a spine loading apparatus. In vitro flexibility testing was performed by applying pure bending moments of 7.5 Nm without and with additional preload of 400 N in the three principal motion planes. Four PDS devices, “DYN” (Dynesys®, Zimmer GmbH, Switzerland), “DSS™” (Paradigm Spine, Wurmlingen, Germany), and two prototypes of dynamic rods, “LSC” with a leaf spring, and “STC” with a spring tube (Aesculap AG, Tuttlingen, Germany), were tested in comparison to a rigid fixation device S⁴ (Aesculap AG, Tuttlingen, Germany) “RIG”, to

the native situation “NAT” and to a defect situation “DEF” of the specimens. The instrumented level was L4–L5. The tested PDS devices comprising a stiffness range for axial stiffness of 10 N/mm to 230 N/mm and for bending stiffness of 3 N/mm to 15 N/mm. Range of motion (ROM), neutral zone (NZ), and intradiscal pressure (IDP) were analyzed for all instrumentation steps and load cases of the instrumented and non-instrumented level. In flexion, extension, and lateral bending, all systems, except STC, showed a significant reduction of ROM and NZ compared to the native situation ($p < 0.05$). Furthermore, we found no significant difference between DYN and RIG ($p > 0.1$). In axial rotation, only DSS and STC reduced the ROM significantly ($p < 0.005$) compared to the native situation, whereas DYN and LSC stayed at the level of the native intersegmental rotation ($p > 0.05$). A correlation was found between axial stiffness and intersegmental stabilization in the sagittal and frontal plane, but not in the transversal plane where intersegmental stabilization is mainly governed by the systems’ ability to withstand shear loads. Furthermore, we observed the systems’ capacity to reduce IDP in the treated segment. The adjacent segment does not seem to be affected by the stiffness of the fixation device under the described loading conditions.

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Keywords Dynamic pedicle screw system · Kinematics ·
Intradiscal pressure · Biomechanics · Spine

Introduction

As a treatment for chronic back pain due to disc degeneration, fusion remains the gold standard [1]. However, this procedure is often accompanied by complications such as accelerated degeneration of the adjacent levels [2–5] and

related to iliac crest bone grafting [6–8]. As alternative treatment, motion-preserving techniques have been developed and introduced, using anterior or posterior devices. The latter, such as interspinous implants or pedicle screw-based dynamic stabilization systems (PDS), are intended to alter the load transfer and the kinematics at the affected level to delay degeneration. However, as of now, it remains unclear when a PDS is clinically indicated and how the ideal PDS mechanism should be designed to achieve this goal.

In clinical use since 1994 [9], Dynesys has shown mixed results, strongly dependent on the clinical indication [10–15]. Apparently, a complication attributed to its inadequate stabilization in the transversal plane may be the disability to eliminate radicular pain [16, 17].

Comparing spinal fusion and PDS devices, several in vitro biomechanical studies found no difference in flexion/extension and lateral bending stabilization, but a clear one in rotatory stabilization, whereas the PDS device was at least similar to the native situation [18–22]. Rohlmann et al. [23] showed that only very low axial stiffness (<200 N/mm) of a posterior rod affects the kinematic response of a segment in the sagittal plane. Higher rod stiffness does not significantly influence the segment kinematics and is comparable to rigid systems. Schmidt et al. and Wilke et al. [24, 25] confirmed these findings in the sagittal plane. They also showed that, to stabilize the treated segment equally in all three motion planes, the bending stiffness of the rod has to be taken into account regarding lateral bending and axial rotation whereas its effect on flexion/extension is negligible.

It is impossible to estimate the overall impact of PDS designs as comparative data are still missing which would allow understanding in more detail the design-related biomechanical consequences. To our knowledge, no in vitro study has yet compared the kinematic response and load transfer within the intervertebral disc (IVD) of different PDS devices with each other and with the golden standard rigid stabilization. Therefore, the aim of this study was to compare different PDS devices against a rigid fixation device to investigate the biomechanical impact of PDS design on stabilization and load transfer in the treated and upper adjacent segment and to gain additional experience for clinical indications and limitations.

Material and method

Specimens

Six fresh-frozen human lumbar spines (L3–L5) with a mean age of 67 years (range 55–77) kept at -21°C in triple sealed bags were thawed overnight at 6°C before the test.

CT scans did not reveal any fractures, osteophytes, or signs of severe disc degeneration. Soft tissue was removed, leaving the ligaments, capsules, and supporting structures intact. In order to fix the specimens firmly in place on the simulator, the cranial and caudal vertebrae (L3 and L5) were embedded with a casting resin (Ureol FC 53, Vantico GmbH, Wehr, Germany) in the test fixtures so that segmental motion was not restricted in any way, and the L3–L4 disc oriented in the horizontal plane.

Instrumentation

Four devices, “DYN” (Dynesys[®], Zimmer GmbH, Switzerland), “DSSTM” (Paradigm Spine, Wurmlingen, Germany), and two prototypes of dynamic rods, “LSC” with a leaf spring, and “STC” with a spring tube (Aesculap AG, Tuttlingen, Germany), were tested in comparison to a rigid fixation device S⁴ (Aesculap AG, Tuttlingen, Germany) “RIG”, to the native situation “NAT” and to a defect situation “DEF” of the lumbar spine (Fig. 1). The instrumented

Fig. 1 Specimens with the different instrumentations: Rigid Fixation (RIG), Spring Tube Concept (STC), Leaf Spring Concept (LSC), DSSTM (DSS), Dynesys (DYN)

level was L4–L5. The segment condition DEF represents a standardized undercutting decompression as described by Schulte et al. [22].

Stiffness

The devices have the following axial stiffness, STC, 10 N/mm; LSC, 70 N/mm; DSS, 50 N/mm; and DYN, 230 N/mm; and the following bending stiffness, STC, 15 N/mm; LSC, 3 N/mm; DSS, 5 N/mm; and DYN, 6 N/mm. The latter was measured in a simple pretest with a sample size of $n = 5$; the one end of the rod is clamped, the other end is subjected to a perpendicular load with a defined lever arm of 30 mm linearly until reaching a deformation of 1.5 mm at the load application point. The measured results showed a high reproducibility with a standard deviation of less than 2% of the mean value. However, it must be noted that it is challenging to evaluate the bending stiffness of dynamic systems as it depends on several factors and overall system design.

Adaptation of DSS and DYN

The systems were adapted to allow the connection with the RIG pedicle screw system. The DSS's connecting elements on the coupler were replaced by the standard Ø 5.5 mm titanium rod used in the RIG system. The DYN was inserted by a tube adaptor, which fits into the RIG screw and has the same cord-locking and PCU spacer mechanism as the original device. The goal was to achieve the same primary clamping stability in the adapted systems as in the original. Subassembly testing of axial rod gripping and flexion bending according to ASTM F1798-97 (2003) brought comparable results. Furthermore, previous synthetic tests [26] had shown that the adaptation had not altered the original range of motion of the systems.

Adaptation of STC and LSC

STC consists of two parts. The basic part comprises a Ø 5.5 mm lower rod and a much smaller upper rod acting as a telescope arm. Its upper counterpart is a Ø 5.5 mm biconical tube on which the telescope arm is mounted, allowing translational and bending movements. A spring connects those two parts. In contrast, LSC is a one-piece design. Both ends are solid Ø 5.5 mm rods with a leaf spring in between. This design allows very linear axial and bending stiffness.

The adaptation of all dynamic rods into a standard pedicle screw made it possible to insert the pedicle screws once, and then perform a cement augmentation through the screw in the vertebra and replace the rods at every instrumentation step (Fig. 2). This procedure minimizes the risk of screw loosening during testing and allows a side-by-side

Fig. 2 X-ray of a bisegmental test specimen (L3–L5) with the cemented pedicle screws as basis platform for the consecutive instrumentation of the different dynamic rod systems

Table 1 Testing order, segment condition and identifier of the used systems

Testing order	Segment condition	Identifier
(1) First	Native	NAT
(2) Second	Defect	DEF
(3) Third	Rigid instrumentation	RIG_1
(4) Random	Spring tube concept	STC
(5) Random	Leaf spring concept	LSC
(6) Random	DSS	DSS
(7) Random	Dynesys	DYN
(8) Last	Rigid instrumentation	RIG_2
Mean of (3) and (8)		RIG_MW

comparison, with the following instrumentation steps designated as segment conditions (Table 1). The instrumentation steps (4)–(7) were randomized for each specimen. The rigid instrumentation was always measured as the first and the last to quantify the effect, if any, of screw loosening or specimen weakening during the entire test period. This also allows capturing the effect of specimen weakening by calculating the mean RIG_MW out of RIG_1 and RIG_2, which can be reasonably compared with the randomized steps (4)–(7).

The test method complies with the testing criteria for spinal implants [27]. The specimens were loaded at room temperature into a spinal simulator based on the principles

of Crawford et al. [28], applying pure moments (± 7.5 Nm) with a velocity of 3°/s for flexion/extension (Flex/Ex), lateral bending (LB) and axial rotation (AR) and in a second step with additional axial preload ($F_P = 400$ N). The kinematics, i.e., the six components of motion according to Panjabi [29] were measured with a 3D ultrasonic motion analysis system (Zebris, Isny, Germany) in the instrumented (L4–L5) and in the upper adjacent segment (L3–L4). The characteristic parameters, range of motion (ROM) and neutral zone (NZ), were analyzed from the hysteresis curves of the third loading cycle.

Intradiscal pressure (IDP) in the treated (L4–L5) and untreated disc (L3–L4) was measured at the same time as the kinematics, using a fiberoptic miniature pressure transducer ($\varnothing$ 0.4 mm, pressure range -0.1 – $1,700$ kPa) based on the Fabry–Perot principle (Samba Sensors, Sweden). The transducer tip was inserted ventro-laterally in the nucleus pulposus with an intravascular indwelling cannula, Introcane® W, $\varnothing$ 0.7 mm (B. Braun, Germany). IDP values were analyzed at three characteristic loading points from the moment versus IDP plots at maximum, zero and minimum moment.

Statistics

The effect of segment condition on ROM, NZ, and IDP as absolute values was assessed using repeated-measures analysis of variance (ANOVA) with a significance level of $p = 0.05$. Prior to analysis, the normal distribution of the data was verified with the Kolmogorov–Smirnov test. Least square difference test (LSD) for post-hoc analysis was used to determine the differences between specific segment conditions. Additionally the correlation between axial rod stiffness and ROM was determined by the Spearman's rank correlation coefficient r_s with a significance level of $p = 0.05$. All statistical analyses were performed with Statistica 8.0 (StatSoft, Inc.).

Results

Range of motion and neutral zone

The ROM and NZ of the treated and untreated segment (L4–L5, L3–L4) for all segment conditions, all principal motion planes, and both load cases are summarized in Tables 2 and 3, shown as absolute values (degrees). The changes of condition in the treated segment in the pure moment mode relative to the native situation are presented in Fig. 3: ROM, and Fig. 4: NZ. Although the DEF increased the ROM in all motion planes, the differences were not significant. The RIG reduced the ROM and NZ significantly in all principal motion planes ($p < 0.0001$).

In flexion, extension, and lateral bending, all systems, except STC, showed a significant reduction of ROM and NZ compared to the native situation ($p < 0.05$). In both loading directions, a correlation was observed between rod stiffness and stabilization with $r_s = -0.77$ for flexion, $r_s = -0.68$ for extension, and $r_s = -0.8$ for lateral bending ($p < 0.0001$). Furthermore, we found no significant difference between DYN and RIG ($p > 0.1$). In addition, the systems with almost the same axial rod stiffness, LSC and DSS, brought about a nearly identical ROM reduction ($p > 0.7$) in flexion, extension, and lateral bending.

In axial rotation, only DSS and STC reduced the ROM significantly ($p < 0.005$) compared to the native situation while DYN and LSC stayed at the level of the native intersegmental rotation ($p > 0.05$). Here, we found no correlation between rod stiffness and stabilization with $r_s = -0.2$ ($p = 0.3$).

Intradiscal pressure

Intradiscal pressure (IDP) results for flexion/extension (Fig. 5) and axial rotation (Fig. 6) are shown as absolute values for all segment conditions and the two loading modes of (a) pure moment and (b) pure moment with additional preload. For lateral bending, we found no consistent data and did not include them in this report. The DEF increased IDP in flexion, extension, and axial rotation. For all instrumentation steps, a further increase of IDP was observed in flexion whereas, in extension, a reduction of IDP below the native situation was found. In axial rotation, IDP was asymmetrical in the treated segment at the maximum torsional moments to the right and left in the pure moment mode. However, the respective segment conditions showed similar trends on both sides. Here, decompression of the segment increased IDP and was most pronounced when rotating to the right side. RIG, STC, LSC, and DSS reduced IDP at least to the native level, whereas the DYN stayed at the level of the pathological situation (Fig. 6a). In the neutral state, compared to the native situation, the averaged change of IDP was 13.2% ($\pm 44.7\%$) for all segment conditions in flexion extension, and axial rotation, in both loading modes.

Additional preload

The additional preload reduced the absolute values of the ROM and NZ in all motion planes and in all segment situations, whereas the change of the different segment conditions relative to the native situation stayed nearly the same. In the neutral state, the preload generally increased IDP. However, the relative change of IDP at the maximum bending moment was less pronounced compared to the pure moment mode.

Table 2 Range of motion (ROM) in (°)—mean $\pm$ standard deviation—for all segment conditions, segment levels, loading directions and load cases

Condition	Flexion	Extension	Flexion/extension	Lateral bending	Axial rotation
Pure moment L4–L5—ROM (°)					
NAT	5.62 ($\pm$ 2.17)	−3.32 ($\pm$ 1.12)	8.94 ($\pm$ 3.13)	7.76 ($\pm$ 1.85)	5.16 ($\pm$ 1.30)
DEF	6.57 ($\pm$ 2.48)	−4.00 ($\pm$ 0.95)	10.57 ($\pm$ 3.22)	8.01 ($\pm$ 1.72)	5.82 ($\pm$ 1.47)
RIG_1	1.10 ($\pm$ 0.32)	−1.22 ($\pm$ 0.41)	2.33 ($\pm$ 0.71)	1.77 ($\pm$ 0.60)	2.03 ($\pm$ 0.44)
STC	3.70 ($\pm$ 1.06)	−2.79 ($\pm$ 0.75)	6.49 ($\pm$ 1.59)	6.37 ($\pm$ 1.58)	3.48 ($\pm$ 0.61)
LSC	2.19 ($\pm$ 0.55)	−1.97 ($\pm$ 0.57)	4.16 ($\pm$ 1.12)	4.46 ($\pm$ 1.26)	4.18 ($\pm$ 1.10)
DSS	2.16 ($\pm$ 0.62)	−1.90 ($\pm$ 0.44)	4.06 ($\pm$ 1.06)	4.17 ($\pm$ 1.11)	3.05 ($\pm$ 0.68)
DYN	1.50 ($\pm$ 0.39)	−1.51 ($\pm$ 0.45)	3.01 ($\pm$ 0.82)	3.12 ($\pm$ 0.98)	4.91 ($\pm$ 1.29)
RIG_2	1.40 ($\pm$ 0.43)	−1.38 ($\pm$ 0.40)	2.78 ($\pm$ 0.82)	2.00 ($\pm$ 0.66)	2.28 ($\pm$ 0.54)
RIG_MW	1.25 ($\pm$ 0.37)	−1.30 ($\pm$ 0.40)	2.55 ($\pm$ 0.76)	1.89 ($\pm$ 0.63)	2.15 ($\pm$ 0.48)
Pure moment L3–L4—ROM (°)					
NAT	4.67 ($\pm$ 1.79)	−2.18 ($\pm$ 0.54)	6.85 ($\pm$ 2.25)	7.66 ($\pm$ 2.91)	4.67 ($\pm$ 2.52)
DEF	5.05 ($\pm$ 2.30)	−2.31 ($\pm$ 0.59)	7.36 ($\pm$ 2.76)	7.53 ($\pm$ 2.94)	4.88 ($\pm$ 2.74)
RIG_1	4.93 ($\pm$ 2.22)	−2.33 ($\pm$ 0.77)	7.26 ($\pm$ 2.88)	7.82 ($\pm$ 3.08)	4.85 ($\pm$ 2.99)
STC	5.43 ($\pm$ 2.61)	−2.20 ($\pm$ 0.53)	7.63 ($\pm$ 3.08)	8.03 ($\pm$ 3.38)	4.90 ($\pm$ 2.85)
LSC	5.38 ($\pm$ 2.61)	−2.26 ($\pm$ 0.81)	7.65 ($\pm$ 2.97)	8.07 ($\pm$ 3.33)	4.94 ($\pm$ 2.89)
DSS	5.48 ($\pm$ 2.69)	−2.04 ($\pm$ 0.43)	7.53 ($\pm$ 3.06)	8.11 ($\pm$ 3.23)	4.89 ($\pm$ 2.92)
DYN	5.37 ($\pm$ 2.57)	−2.28 ($\pm$ 0.65)	7.65 ($\pm$ 3.11)	7.95 ($\pm$ 3.34)	5.15 ($\pm$ 3.01)
RIG_2	5.68 ($\pm$ 2.86)	−2.19 ($\pm$ 0.56)	7.86 ($\pm$ 3.37)	8.22 ($\pm$ 3.60)	4.89 ($\pm$ 2.71)
RIG_MW	5.30 ($\pm$ 2.53)	−2.26 ($\pm$ 0.66)	7.56 ($\pm$ 3.12)	8.02 ($\pm$ 3.34)	4.87 ($\pm$ 2.84)
Pure moment and preload L4–L5—ROM (°)					
NAT	5.78 ($\pm$ 2.44)	−2.50 ($\pm$ 1.37)	8.28 ($\pm$ 3.30)	5.68 ($\pm$ 2.20)	3.82 ($\pm$ 1.00)
DEF	6.87 ($\pm$ 2.61)	−2.84 ($\pm$ 1.53)	9.72 ($\pm$ 3.48)	5.70 ($\pm$ 2.20)	4.16 ($\pm$ 1.17)
RIG_1	1.00 ($\pm$ 0.32)	−0.99 ($\pm$ 0.27)	1.99 ($\pm$ 0.58)	1.48 ($\pm$ 0.55)	1.75 ($\pm$ 0.32)
STC	3.27 ($\pm$ 0.92)	−2.04 ($\pm$ 0.51)	5.31 ($\pm$ 1.20)	4.05 ($\pm$ 1.59)	2.16 ($\pm$ 0.39)
LSC	2.04 ($\pm$ 0.52)	−1.80 ($\pm$ 0.45)	3.84 ($\pm$ 0.96)	3.14 ($\pm$ 1.18)	3.22 ($\pm$ 0.82)
DSS	2.04 ($\pm$ 0.48)	−1.70 ($\pm$ 0.40)	3.74 ($\pm$ 0.86)	3.08 ($\pm$ 1.09)	2.42 ($\pm$ 0.55)
DYN	1.36 ($\pm$ 0.38)	−1.28 ($\pm$ 0.36)	2.64 ($\pm$ 0.72)	2.33 ($\pm$ 0.94)	3.86 ($\pm$ 1.05)
RIG_2	1.19 ($\pm$ 0.36)	−1.12 ($\pm$ 0.36)	2.31 ($\pm$ 0.69)	1.68 ($\pm$ 0.66)	1.95 ($\pm$ 0.44)
RIG_MW	1.09 ($\pm$ 0.34)	−1.05 ($\pm$ 0.31)	2.15 ($\pm$ 0.63)	1.58 ($\pm$ 0.61)	1.85 ($\pm$ 0.38)
Pure moment and preload L3–L4—ROM (°)					
NAT	5.41 ($\pm$ 2.38)	−1.73 ($\pm$ 0.61)	7.14 ($\pm$ 2.71)	6.71 ($\pm$ 3.40)	3.68 ($\pm$ 2.39)
DEF	5.54 ($\pm$ 2.54)	−1.83 ($\pm$ 0.89)	7.37 ($\pm$ 2.95)	6.65 ($\pm$ 3.31)	3.54 ($\pm$ 2.37)
RIG_1	5.48 ($\pm$ 2.72)	−1.69 ($\pm$ 0.54)	7.17 ($\pm$ 3.04)	6.72 ($\pm$ 3.43)	3.38 ($\pm$ 2.45)
STC	5.86 ($\pm$ 3.02)	−1.64 ($\pm$ 0.68)	7.50 ($\pm$ 3.22)	6.88 ($\pm$ 3.54)	3.28 ($\pm$ 2.33)
LSC	5.83 ($\pm$ 2.91)	−1.65 ($\pm$ 0.60)	7.49 ($\pm$ 3.16)	6.78 ($\pm$ 3.52)	3.40 ($\pm$ 2.27)
DSS	5.70 ($\pm$ 2.81)	−1.65 ($\pm$ 0.63)	7.35 ($\pm$ 3.13)	6.95 ($\pm$ 3.51)	3.38 ($\pm$ 2.39)
DYN	5.71 ($\pm$ 2.85)	−1.62 ($\pm$ 0.54)	7.34 ($\pm$ 3.22)	6.72 ($\pm$ 3.64)	3.46 ($\pm$ 2.31)
RIG_2	5.76 ($\pm$ 3.01)	−1.72 ($\pm$ 0.86)	7.49 ($\pm$ 3.47)	6.95 ($\pm$ 3.80)	3.27 ($\pm$ 2.27)
RIG_MW	5.62 ($\pm$ 2.86)	−1.70 ($\pm$ 0.69)	7.33 ($\pm$ 3.26)	6.83 ($\pm$ 3.61)	3.32 ($\pm$ 2.36)

Abbreviations are explained in Table 1

Untreated segment

In the untreated adjacent segment, we observed minor changes in ROM and NZ in all the three motion planes, in

both loading scenarios, and for all segment conditions compared to the native situation. In the pure moment mode, the changes in ROM were 8.3% ($\pm$ 12.5%) in flexion/extension, 2.9% ($\pm$ 5.4%) in lateral bending, and 2.6%

Table 3 Neutral Zone (NZ) in (°)—mean $\pm$ standard deviation—for all segment conditions, segment levels, loading directions and load cases

Condition	Flexion/extension	Lateral bending	Axial rotation
Pure moment L4–L5—NZ (°)			
NAT	4.24 ($\pm$ 1.26)	4.02 ($\pm$ 0.90)	2.14 ($\pm$ 0.33)
DEF	5.10 ($\pm$ 1.10)	4.06 ($\pm$ 0.97)	2.21 ($\pm$ 0.50)
RIG_1	1.13 ($\pm$ 0.27)	0.72 ($\pm$ 0.25)	0.79 ($\pm$ 0.23)
STC	3.55 ($\pm$ 0.86)	2.78 ($\pm$ 1.31)	1.50 ($\pm$ 0.25)
LSC	1.73 ($\pm$ 0.32)	1.47 ($\pm$ 0.26)	1.60 ($\pm$ 0.41)
DSS	1.83 ($\pm$ 0.30)	1.50 ($\pm$ 0.30)	1.40 ($\pm$ 0.09)
DYN	1.45 ($\pm$ 0.32)	1.19 ($\pm$ 0.28)	2.72 ($\pm$ 0.84)
RIG_2	1.41 ($\pm$ 0.34)	0.79 ($\pm$ 0.25)	0.89 ($\pm$ 0.26)
RIG_MW	1.27 ($\pm$ 0.30)	0.75 ($\pm$ 0.25)	0.84 ($\pm$ 0.25)
Pure moment L3–L4—NZ (°)			
NAT	2.03 ($\pm$ 0.48)	2.75 ($\pm$ 0.53)	1.67 ($\pm$ 0.81)
DEF	1.91 ($\pm$ 0.57)	2.71 ($\pm$ 0.64)	1.46 ($\pm$ 0.56)
RIG_1	2.70 ($\pm$ 0.81)	3.06 ($\pm$ 0.70)	2.13 ($\pm$ 0.98)
STC	2.43 ($\pm$ 0.55)	2.86 ($\pm$ 1.29)	2.11 ($\pm$ 1.17)
LSC	2.71 ($\pm$ 0.95)	3.10 ($\pm$ 0.79)	2.20 ($\pm$ 1.17)
DSS	2.41 ($\pm$ 0.40)	3.25 ($\pm$ 0.76)	1.84 ($\pm$ 0.88)
DYN	2.74 ($\pm$ 0.68)	3.20 ($\pm$ 1.03)	2.17 ($\pm$ 1.36)
RIG_2	2.80 ($\pm$ 0.75)	3.80 ($\pm$ 1.49)	2.18 ($\pm$ 0.99)
RIG_MW	2.75 ($\pm$ 0.78)	3.43 ($\pm$ 1.05)	2.16 ($\pm$ 0.99)
Pure moment and preload L4–L5—NZ (°)			
NAT	2.39 ($\pm$ 0.64)	1.99 ($\pm$ 0.82)	1.60 ($\pm$ 0.21)
DEF	2.55 ($\pm$ 0.73)	1.89 ($\pm$ 0.67)	1.64 ($\pm$ 0.38)
RIG_1	0.90 ($\pm$ 0.23)	0.56 ($\pm$ 0.21)	0.84 ($\pm$ 0.18)
STC	2.38 ($\pm$ 0.82)	2.06 ($\pm$ 0.75)	1.05 ($\pm$ 0.20)
LSC	1.40 ($\pm$ 0.26)	1.09 ($\pm$ 0.34)	1.43 ($\pm$ 0.38)
DSS	1.47 ($\pm$ 0.26)	1.06 ($\pm$ 0.38)	1.10 ($\pm$ 0.26)
DYN	1.15 ($\pm$ 0.24)	0.93 ($\pm$ 0.34)	1.95 ($\pm$ 0.57)
RIG_2	1.01 ($\pm$ 0.26)	0.60 ($\pm$ 0.22)	0.91 ($\pm$ 0.23)
RIG_MW	0.96 ($\pm$ 0.23)	0.58 ($\pm$ 0.21)	0.88 ($\pm$ 0.20)
Pure moment and preload L3–L4—NZ (°)			
NAT	1.38 ($\pm$ 0.49)	1.68 ($\pm$ 0.44)	1.40 ($\pm$ 0.68)
DEF	1.28 ($\pm$ 0.51)	1.71 ($\pm$ 0.42)	1.38 ($\pm$ 0.77)
RIG_1	1.91 ($\pm$ 0.37)	2.14 ($\pm$ 0.45)	1.51 ($\pm$ 0.81)
STC	1.57 ($\pm$ 0.43)	1.81 ($\pm$ 0.31)	1.41 ($\pm$ 0.75)
LSC	1.84 ($\pm$ 0.40)	1.97 ($\pm$ 0.52)	1.49 ($\pm$ 0.85)
DSS	1.82 ($\pm$ 0.41)	2.02 ($\pm$ 0.39)	1.43 ($\pm$ 0.79)
DYN	1.88 ($\pm$ 0.44)	2.12 ($\pm$ 0.58)	1.45 ($\pm$ 0.91)
RIG_2	2.02 ($\pm$ 0.48)	2.08 ($\pm$ 0.48)	1.39 ($\pm$ 0.71)
RIG_MW	1.96 ($\pm$ 0.41)	2.11 ($\pm$ 0.44)	1.45 ($\pm$ 0.76)

Abbreviations are explained in Table 1

($\pm$ 10%) in axial rotation, whereas changes in NZ were 24.6% ($\pm$ 31.3%), 11.9% ($\pm$ 20.6%) and 17.6% ($\pm$ 29.2%), respectively.

In addition, in the untreated segment for all segment conditions compared to the native situation, the change of

IDP averaged 11.5% ($\pm$ 36.8%) in flexion/extension and axial rotation and in both loading modes.

Discussion

A number of concepts with controversy approaches are currently discussed in dynamic pedicle screw systems. However, a direct comparison among the different systems has so far not been performed. This study investigated the effect of different dynamic pedicle screw systems on the kinematic response and load transfer within the treated (L4–L5) and untreated adjacent segment (L3–L4). The protocol was designed to allow a direct side-by-side comparison of dorsal instrumentations in the same specimen as only the rods were changed. To this effect, we inserted a standard pedicle screw, and then performed a cement augmentation through the screw to minimize loosening during testing. For this reason, the current study is, to our knowledge, the first to provide a direct comparison of such systems in the same specimen.

In the dynamic instrumentations, we found a correlation between axial stiffness and stabilization in the sagittal and frontal plane. Here, the system with the lowest stiffness, STC, displayed the lowest stabilization of the treated segment, whereas the system with the highest stiffness, DYN, was most pronounced and not significantly different from rigid instrumentation. The latter reduced the ROM and the NZ significantly in all the three motion planes as expected. In the transversal plane, the systems behaved differently that is we did not find any correlation between stiffness and stabilization. The system's ability to stabilize the segment, that is its ability to withstand shear loads, depends mainly on design and not, as assumed, on axial or bending stiffness.

The native ROM of the functional spinal units tested in the present study was, in all motion planes, comparable to the values reported in the literature for in vitro testing of specimens in the loading modes of pure moment [20, 21, 25, 30, 31] and pure moment with additional preload [19, 20]. In addition, the ROM data reported here on DYN are consistent with comparable studies, where DYN shows nearly the same stabilizing effect as a rigid fixation device in flexion/extension and lateral bending, but very limited stabilization in axial rotation [18–22, 30]. In addition, the data on DSS are also consistent with the findings of Wilke et al. [25], except for axial rotation. They reported a ROM reduction of 54% in flexion, 39% in extension, 45% in lateral bending, and 7% in axial rotation in comparison to our results of 58.2, 40.4, 46.3, and 52.5%, respectively. The differences in axial rotation could be due to screw loosening, as described by Schmidt et al. [24], as the rigid fixation system they tested showed only a decrease of 26%,

Fig. 3 Range of motion (ROM) of the treated L4–L5 segment. The different segment conditions are normalized to the native situation representing 100%. Load case: pure moment. Abbreviations are explained in Table 1

which is in the lower range compared to other studies [21, 32–36].

FEA studies have shown that rod stiffness plays a major role in stabilizing the segment after treatment with a dynamic rod system. Here, Rohmann et al. [23] reported that only very low rod stiffness <200 N/mm markedly influences intersegmental rotation at the treated level in flexion/extension. In a more detailed study on rod, axial and bending stiffness, and on its influence on intersegmental rotation, Schmidt et al. [24] demonstrated a correlation between implant stiffness and segment stabilization in all the three main motion planes. In contrast, in flexion/extension, the effect of bending stiffness was negligible.

Our results confirm these findings on flexion/extension. However, they also show this clear correlation between axial stiffness and segment stabilization in lateral bending as in flexion and extension, and the same negligible effect of bending stiffness. Furthermore, on axial rotation, our results do not support the hypothesis that bending and axial stiffness of the rod are the only factors in achieving a

certain stabilization of the treated segment, but they suggest that the rod design itself plays a major role in this context, as shown for the different system designs used in this study with a range of axial stiffness between 10 and 230 N/mm and almost the same bending stiffness of below 15 N/mm in the first linear range. If axial stiffness were the predominant factor, the system with the lowest stiffness would have the smallest stabilizing effect. Nevertheless, as we have seen, there is no correlation between axial stiffness and stabilizing effect. Therefore, DYN with the highest stiffness had the smallest stabilizing effect, whereas STC, the system with the lowest stiffness, had a median stabilizing effect compared to DSS and LSC, the two systems with the same stiffness. This could be due to the specific shear stiffness of the different systems. The DYN spacer is locked into place by a cord between the flat sides of the screw body, thus it cannot withstand high shear load and could shift transversely at the screw ends. The one-piece design of LSC also has low shear stiffness and allows translation in anterior-posterior and medial–lateral

Fig. 4 Neutral Zone (NZ) of the treated L4–L5 segment. The different segment conditions are normalized to the native situation representing 100%. Load case: pure moment. Abbreviations are explained in Table 1

direction. In the system designs using a compression spring mechanism (STC, DSS), shear resistance is obviously higher because translational movement is more restricted. In these two systems, we found a correlation between axial

stiffness and stabilization. Therefore, our findings suggest that a system's shear stiffness is the critical factor in achieving stabilization in the transversal plane.

IDP data showed an increase in the sagittal and transversal plane after decompression of the treated segment. After instrumentation with RIG, we observed a further increase of IDP in flexion, but a clear reduction in extension. The dynamic systems demonstrated a similar trend, whereas STC showed the smallest increase of IDP in flexion, but also the smallest reduction in extension. The same trend was found in the second loading mode with additional preload. In the transversal plane, the RIG, STC, LSC, and DSS instrumentation steps reduced IDP in the treated segment with the exception of the DYN, which stayed at the level of the decompressed segment condition.

These pressure values match those obtained by Schmoelz et al. [37]. In the native situation, IDP was found in the same range in all motion planes; this also applies to the effect of decompression and instrumentation with a rigid fixator and the Dynesys system on IDP.

In contrast, IDP remained at the same level in the unloaded neutral condition of the segments in the different instrumentation steps while the instrumentation steps of the rigid fixator and the Dynesys system had a different effect. Our results show a trend towards a further increase for all instrumentations in flexion. Apparently, the center of rotation shifts to the dorsal structures after posterior instrumentation leading to a higher loading of the IVD in flexion due to the pre-tensioning effect of the systems.

In the preload mode, IDP values increased clearly, but the changes were smaller at maximum moment. This means that the additional preload causes a more constant IDP during a complete loading cycle. However, certain effects of the instrumentations are still present.

These pressure values approximate those recorded in certain postures by Nachemson [38], Sato [39], and Wilke [40]. Thus, pure moment loading creates the IDP measured in a lying position and preloading that measured in an upright position.

The IDP data also support the hypothesis that design parameters, such as implant stiffness, result in a different load transfer within the intervertebral disc. Thus, the system with the lowest axial stiffness brought about the smallest decrease of IDP in extension, but also the smallest increase in flexion, and the rigid instrumentation the highest decrease in extension, but also the highest increase in flexion.

Furthermore, in axial rotation, design was seen to influence IDP. Here, DYN did not reduce IDP in comparison to decompression, but the RIG did so to the highest degree.

Data on lateral bending were not consistent and very asymmetrical, possibly due to the medio-lateral position of

Fig. 5 Intradiscal Pressure (IDP) for all segment conditions as absolute values (kPa) of the treated segment (L4–L5) at maximum moment (*Flexion*), zero moment (*Neutral*) and minimum moment (*Extension*). Load cases: **a** pure moment, **b** pure moment and additional preload

Fig. 6 Intradiscal Pressure (IDP) for all segment conditions as absolute values (kPa) of the treated segment (L4–L5) at maximum moment (*AR, right*), zero moment (*Neutral*) and minimum moment (*AR, left*). Load cases: **a** pure moment, **b** pure moment and additional preload

the sensor. Extensive pretest series were conducted for the development and evaluation of the methodology to measure the intradiscal pressure with a high reproducibility and with the goal to minimize artifacts due to sensor positioning or placement. From this pretest series, the medio-lateral sensor position was found to be optimal for this kind of experimental procedure. Despite of the good pretest results, the pressure measurement for lateral bending conducted in the here reported study was found to be very asymmetrical. However, to be consistent with the outlined method a change of the sensor position during the test was not considered. For this reason, we did not include this data in this report.

In the untreated segment, we observed no changes of ROM, NZ, and IDP in all treatment conditions. However, in vitro testing is clearly limited in its ability to investigate the influence of instrumentation techniques in the adjacent

level. The pure moment protocol for flexibility testing used in this study causes constant moment loading at all segmental levels, making a change in ROM improbable. Additional loading with superimposed preload probably alters this behavior because of load vectors acting in a different way, but did not change the ROM in all systems significantly. We made the hitherto unreported observation that the NZ of the adjacent segment increased by trend after both rigid and dynamic instrumentation, but not in the pathological situation. We found no conclusive explanation for this fact, but it may be due to compensating effects increasing the laxity of the adjacent segment.

The used specimens were from fairly old donors, bringing the risk of degeneration especially of the IVD. Prior to testing they were CT scanned and visually inspected for signs of severe disc degeneration and excluded if such were found. We were aware that the age of the

specimens and the accompanied age related disc degeneration was not ideal to collect reliable data on IDP, as the large standard deviation in the IDP measurements shows. Nevertheless, we did not think it would have a significant impact on the kinematic response. The PDS systems are generally implanted in younger patients, but they are intended to treat and correct problems caused by disc degeneration. In addition, previous studies have shown that degeneration of the IVD has only a minimal effect on the ROM [41] and may even result in a decrease in a severe case [31].

To test five systems on the same specimen could be seen as a limitation of the current study as screw loosening could occur. In order to address this issue, we used a standard pedicle screw system inserted by cement augmentation. In addition, to eliminate this distorting effect, we repeated the rigid instrumentation (RIG_1) after the dynamic instrumentations as last step (RIG_2). The ROM increased by no more than 5% compared to the native ROM, as in the untreated adjacent segment (Table 2). For this procedure, the two dynamic systems, DSS and DYN, had to be adapted to the one standard pedicle screw system. Pretests confirmed that this adaptation did not alter their original mechanical behavior [26].

In addition, we inserted the original DYN PCU spacer used in clinical praxis. Several authors suggest using a special PCU spacer for in vitro testing with a lower stiffness than that of the original because temperature affects spacer stiffness. A pretest yielded a stiffness of 230 N/mm for the PCU spacer at room temperature, a value acceptably close to the 200 N/mm recorded for the clinically used spacer. The use of such a special spacer is problematic as the cord is subjected to 300 N pre-tensioning, which could cause a compression of the treated level and lead to misleading ROM results for the system in question. This is in agreement with the findings from Niosi et al. [20] who used such a modified spacer. They assume that the longer spacer (+2 mm) better mimics the in vivo situation. For this reason, we decided to use the slightly stiffer spacer; it also added to the range of stiffness magnitudes for the systems included in the study.

Clinical data for posterior pedicle screw-based dynamic stabilization show mixed outcomes ranging from comparable to fusion [10, 11, 13–15, 42] to worse than fusion [16, 17], which could be attributed to the different clinical indications. Obviously, the latter (e.g., disc degeneration, spinal stenosis with or without degenerative spondylolisthesis, etc.) together with the clinical procedure (e.g., with or without decompression, nucleotomy, etc.) is critical for the outcome, but the ability of the systems to restrict or allow movement to a certain degree is also a factor (e.g., if the pain comes from rotation, but the dynamic system does not restrict it, radicular pain may persist).

Although it remains unclear which stiffness or design will prove the most successful for dynamic stabilization of the spine, we believe that it is essential that to restrict the movement of the treated segment in all motion planes to a certain degree, inferior to that of the native situation, to compensate for example the ROM increase from decompression and to minimize abnormal movement or instability.

Conclusion

A correlation was found between axial stiffness and intersegmental stabilization in the sagittal and frontal plane. There the system with the lowest stiffness displayed the lowest stabilization of the treated segment, whereas the system with the highest stiffness was most pronounced and not significantly different from rigid instrumentation. However, no correlation was found between axial stiffness and intersegmental stabilization in the transversal plane where intersegmental stabilization is mainly governed by the systems' ability to withstand shear loads. Furthermore, we observed the systems' capacity to reduce IDP in the treated segment. Our results do not suggest that the adjacent segment is affected by the stiffness of the fixation device under the described loading conditions.

References

1. Fritzell P, Hagg O, Wessberg P, Nordwall A (2001) 2001 Volvo award winner in clinical studies: lumbar fusion versus nonsurgical treatment for chronic low back pain: a multicenter randomized controlled trial from the Swedish Lumbar Spine Study Group. *Spine* 26(23):2521–2532
2. Eck JC, Humphreys SC, Hodges SD (1999) Adjacent-segment degeneration after lumbar fusion: a review of clinical, biomechanical, and radiologic studies. *Am J Orthop* 28(6):336–340
3. Lee CK (1988) Accelerated degeneration of the segment adjacent to a lumbar fusion. *Spine* 13(3):375–377
4. Lehmann TR, Spratt KF, Tozzi JE et al (1987) Long-term follow-up of lower lumbar fusion patients. *Spine* 12(2):97–104
5. Schlegel JD, Smith JA, Schleusener RL (1996) Lumbar motion segment pathology adjacent to thoracolumbar, lumbar, and lumbosacral fusions. *Spine* 21(8):970–981
6. Younger EM, Chapman MW (1989) Morbidity at bone graft donor sites. *J Orthop Trauma* 3(3):192–195
7. Banwart JC, Asher MA, Hassanein RS (1995) Iliac crest bone graft harvest donor site morbidity. A statistical evaluation. *Spine* 20(9):1055–1060
8. Wippermann BW, Schratz HE, Steeg S, Tschern H (1997) [Complications of spongiosa harvesting of the iliac crest. A retrospective analysis of 1,191 cases] (Komplikationen der Spongiosaentnahme am Beckenkamm. Eine retrospektive Analyse von 1191 Fallen.). *Der Chirurg: Zeitschrift für alle Gebiete der operativen Medizin* 68(12):1286–1291
9. Dubois G, de Gernay B, Schaerer NS, Fennema P (1999) Dynamic Neutralization: a new concept for restabilization of the spine, pp 233–240

10. Cakir B, Ulmar B, Koeppe H et al (2003) (Posterior dynamic stabilization as an alternative for dorso-ventral fusion in spinal stenosis with degenerative instability) Dorsale dynamische Stabilisierung als Alternative zur dorso-ventralen Fusion bei Spinalkanalstenose mit degenerativer Instabilität. *Zeitschrift für Orthopädie und ihre Grenzgebiete* 141(4):418–424
11. Putzier M, Schneider SV, Funk J, Perka C (2004) (Application of a dynamic pedicle screw system (DYNESYS) for lumbar segmental degenerations—comparison of clinical and radiological results for different indications) Die Anwendung eines dynamischen Pedikelschraubensystems (DYNESYS) bei lumbalen Segmentdegenerationen: ein Vergleich klinischer und radiologischer Ergebnisse bei unterschiedlichen Indikationen. *Zeitschrift für Orthopädie und ihre Grenzgebiete* 142(2):166–173
12. Putzier M, Schneider SV, Funk JF, Tohtz SW, Perka C (2005) The surgical treatment of the lumbar disc prolapse: nucleotomy with additional transpedicular dynamic stabilization versus nucleotomy alone. *Spine* 30(5):E109–E114
13. Schnake KJ, Schaeren S, Jeanneret B (2006) Dynamic stabilization in addition to decompression for lumbar spinal stenosis with degenerative spondylolisthesis. *Spine* 31(4):442–449
14. Stoll TM, Dubois G, Schwarzenbach O (2002) The dynamic neutralization system for the spine: a multi-center study of a novel non-fusion system. *Eur Spine J* 11(Suppl 2):S170–S178
15. Welch WC, Cheng BC, Awad TE et al (2007) Clinical outcomes of the Dynesys dynamic neutralization system: 1-year preliminary results. *Neurosurg Focus* 22(1):E8
16. Grob D, Benini A, Junge A, Mannion AF (2005) Clinical experience with the Dynesys semirigid fixation system for the lumbar spine: surgical and patient-oriented outcome in 50 cases after an average of 2 years. *Spine* 30(3):324–331
17. Wurgler-Hauri CC, Kalbarczyk A, Wiesli M, Landolt H, Fandino J (2008) Dynamic neutralization of the lumbar spine after microsurgical decompression in acquired lumbar spinal stenosis and segmental instability. *Spine* 33(3):E66–E72
18. Cheng BC, Gordon J, Cheng J, Welch WC (2007) Immediate biomechanical effects of lumbar posterior dynamic stabilization above a circumferential fusion. *Spine* 32(23):2551–2557
19. Gedet P, Haschtmann D, Thistlethwaite PA, Ferguson SJ (2009) Comparative biomechanical investigation of a modular dynamic lumbar stabilization system and the Dynesys system. *Eur Spine J* 18(10):1504–1511
20. Niosi CA, Zhu QA, Wilson DC et al (2006) Biomechanical characterization of the three-dimensional kinematic behaviour of the Dynesys dynamic stabilization system: an in vitro study. *Eur Spine J* 15(6):913–922
21. Schmoelz W, Huber JF, Nydegger T et al (2003) Dynamic stabilization of the lumbar spine and its effects on adjacent segments: an in vitro experiment. *J Spinal Disord Tech* 16(4):418–423
22. Schulte TL, Hurschler C, Haversath M et al (2008) The effect of dynamic, semi-rigid implants on the range of motion of lumbar motion segments after decompression. *Eur Spine J* 17(8):1057–1065
23. Rohlmann A, Burra NK, Zander T, Bergmann G (2007) Comparison of the effects of bilateral posterior dynamic and rigid fixation devices on the loads in the lumbar spine: a finite element analysis. *Eur Spine J* 16(8):1223–1231
24. Schmidt H, Heuer F, Wilke HJ (2009) Which axial and bending stiffnesses of posterior implants are required to design a flexible lumbar stabilization system? *J Biomech* 42(1):48–54
25. Wilke HJ, Heuer F, Schmidt H (2009) Prospective design delineation and subsequent in vitro evaluation of a new posterior dynamic stabilization system. *Spine* 34(3):255–261
26. Schilling C, Vieweg U, Grupp TM, Bloemer W (2008) A synthetic in vitro model for a standardized comparison of dynamic stabilization devices. *J Biomech* 41(S1):1
27. Wilke HJ, Wenger K, Claes L (1998) Testing criteria for spinal implants: recommendations for the standardization of in vitro stability testing of spinal implants. *Eur Spine J* 7(2):148–154
28. Crawford NR, Brantley AG, Dickman CA, Koeneman EJ (1995) An apparatus for applying pure nonconstraining moments to spine segments in vitro. *Spine* 20(19):2097–2100
29. Panjabi MM, Oxland TR, Yamamoto I, Crisco JJ (1994) Mechanical behavior of the human lumbar and lumbosacral spine as shown by three-dimensional load-displacement curves. *J Bone Joint Surg Am* 76(3):413–424
30. Freudiger S, Dubois G, Lorrain M (1999) Dynamic neutralisation of the lumbar spine confirmed on a new lumbar spine simulator in vitro. *Arch Orthop Trauma Surg* 119(3–4):127–132
31. Fujiwara A, Lim TH, An HS et al (2000) The effect of disc degeneration and facet joint osteoarthritis on the segmental flexibility of the lumbar spine. *Spine* 25(23):3036–3044
32. Arand M, Wilke HJ, Schultheiss M et al. (2000) Comparative stability of the “Internal Fixator” and the “Universal Spine System” and the effect of crosslinking transfixating systems. A biomechanical in vitro study. *Biomed Tech (Berl)* 45(11):311–316
33. Eger W, Kluger P, Claes L, Wilke HJ (1999) Characteristics of an extended internal fixation system for polysegmental transpedicular reduction and stabilization of the thoracic, lumbar, and lumbosacral spine. *Eur Spine J* 8(1):61–69
34. Panjabi MM, Henderson G, James Y, Timm JP (2007) StabilimaxNZ) versus simulated fusion: evaluation of adjacent-level effects. *Eur Spine J* 16(12):2159–2165
35. Quint U, Wilke HJ, Loer F, Claes L (1998) [Possibilities for static and dynamic stabilization of the spine in lesions of the anterior and posterior ligament complex] (Statische und dynamische Stabilisierungsmöglichkeiten der Wirbelsäule bei Läsionen des anterioren und posterioren Ligamentkomplexes.). *Der Unfallchirurg* 101(9):684–690
36. Schmoelz W, Onder U, Martin A, von SA (2009) Non-fusion instrumentation of the lumbar spine with a hinged pedicle screw rod system: an in vitro experiment. *Eur Spine J* 18(10):1478–1485
37. Schmoelz W, Huber JF, Nydegger T, Claes L, Wilke HJ (2006) Influence of a dynamic stabilisation system on load bearing of a bridged disc: an in vitro study of intradiscal pressure. *Eur Spine J* 15(8):1276–1285
38. Nachemson AL (1981) Disc pressure measurements. *Spine* 6(1):93–97
39. Sato K, Kikuchi S, Yonezawa T (1999) In vivo intradiscal pressure measurement in healthy individuals and in patients with ongoing back problems. *Spine* 24(23):2468–2474
40. Wilke HJ, Neef P, Caimi M, Hoogland T, Claes LE (1999) New in vivo measurements of pressures in the intervertebral disc in daily life. *Spine* 24(8):755–762
41. Mimura M, Panjabi MM, Oxland TR et al (1994) Disc degeneration affects the multidirectional flexibility of the lumbar spine. *Spine* 19(12):1371–1380
42. Sapkas GS, Themistocleous GS, Mavrogenis AF et al (2007) Stabilization of the lumbar spine using the dynamic neutralization system. *Orthopedics* 30(10):859–865