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Three-dimensional stiffness in a thoracolumbar en-bloc spondylectomy model: A biomechanical in vitro study

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Abstract

Background. In selected cases, en-bloc spondylectomy is the only option to reach wide resection margins for patients with malignant tumours of the thoracolumbar spine. These patients must be also provided a secure initial stabilization of the spine and this is the role of vertebral body replacements employed with posterior fixation systems. The aim of this study was to determine the postimplantation stiffness of a connected vertebral body replacement pedicle screw system in different implantation scenarios following an en-bloc spondylectomy. Reconstruction was varied by posterior fixation lengths and axial compression forces during implantation.

Methods. Three-dimensional stiffness was assessed in 6 fresh frozen human spinal specimens (Th11-L3) using a six degree of freedom spine simulator. Following en-bloc spondylectomy reconstruction was performed using a carbon composite fibre vertebral body replacement connected to a posterior fixation system by two artificial pedicles. The spines were loaded with pure moments (7.5 Nm) in the three main motion planes. The intersegmental rotations were measured between Th12 and L2.

Findings. Reconstructions using long posterior fixation modes demonstrated significant ($P < 0.05$) higher stiffness compared to short posterior fixations in all motion planes. In axial rotation short posterior fixation modes failed to reach the values of the intact state. Neither high nor low axial compression force during implantation showed a significant impact on postfusional stiffness.

Interpretation. In this biomechanical model, the employed system should be implanted with a posterior fixation of two adjacent segments to the lesion in order to achieve a secure stabilization of the treated segment.

Keywords: En-bloc spondylectomy; Biomechanical; In vitro study; Three-dimensional stability

1. Introduction

When treating patients with solitary spinal malignancies en-bloc spondylectomy is performed in curative intention to provide reduced local recurrence rates and to improve overall outcome. Advanced and demanding techniques for en-bloc excisions were published by various authors (Roy-Camille et al., 1981; Stener, 1971; Tomita et al., 1994) which aim to attain secure resection margins and to prevent intraoperative tumour cell dissemination. Classic corporectomy in a piecemeal fashion leaves the dorsal column intact providing up to 40% of residual segmental stability (Gradl, 2006; James et al., 1994). To the contrary, en-bloc resection of all bony and ligamentous vertebral structures leads to complete loss of spinal continuity and stability (Ebihara et al., 2004) leaving neurovascular structures unprotected. Therefore subsequent secure and stable spinal reconstruction is absolutely required. In contrast to degenerative or posttraumatic diseases physiological healing, ingrowth and complete bony fusion can not be

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expected for tumour patients. The postoperative course can be complicated by prolonged instability as a result of progressive implant loosening, hardware failure and delayed or disturbed healing processes, which is caused by patient's poor general health status and concomitant local radiation- and/or chemotherapies (Bouchard et al., 1994). Ghogawala et al. (2001) showed in 2001, a threefold higher risk of postoperative wound healing disturbances in patients that underwent radiation prior to surgery compared to a de novo surgery control group.

In order to reach sufficient postoperative spinal stability, to restore spinal function and to ensure the later healing process (Akamaru et al., 2005) the implant has to provide a stable posterior fixation and anchorage of the vertebral body replacement (VBR) (Schreiber et al., 2005; Shannon et al., 2004). Therefore, precise knowledge of the biomechanical stiffness and behaviour of the employed implant system and methods is an essential precondition for safe clinical application and long term function of the implant system. To date, however, there is no adequate biomechanical model that allows both, a surgical replication of a segmental defect after en-bloc excision as well as a quantitative biomechanical analysis of the resulting reconstruction.

Implantation failures followed by implant dislocation (Gradl, 2006; Knop et al., 2002) or subsidence (Hollowell et al., 1996) are often caused by a lack of standardized implantation procedures. From a clinical perspective, the lack of relevant biomechanical information means that anterior and posterior en-bloc spondylectomy modes (type of VBR, mode of implant fixation, application of axial compression forces, extension of posterior fixation etc.) are performed mainly on empirical evidence as opposed to clear cut biomechanical data. While several biomechanical experiments have shown the need for additional posterior fixation to improve construct stability in en-bloc spondylectomy and corporectomy models (Oda et al., 1999; Vahldiek and Panjabi, 1998), several factors remain unclear. The influence of the length of a posterior fixation (fixation with pedicle screws and rods) with a different number of stabilized adjacent segments remains controversial (Burney et al., 2005; Eichholz et al., 2004). The fit of a vertebral body replacement system is expected to have a distinct effect (Thongtrangan et al., 2003) on initial stability of anterior reconstructions. This depends on a variety of factors, such as the compression forces applied to the construct during implantation (Schultheiss et al., 2003), the surface of osseous tissue (Knoller et al., 2005), the resulting interface contact (Hollowell et al., 1996; Knoller et al., 2005; Wu et al., 1998) and to a great extent on patients mobility in subsequent everyday life (Akamaru et al., 2005).

Therefore, the present in vitro study aimed to quantitatively analyse the biomechanical features of a connected carbon composite/titanium VBR pedicle screw system. Detailed objectives were to investigate the effect of two posterior fixation lengths and two axial compression force options during implantation on primary stiffness in a defect model (en-bloc spondylectomy) of the human thoracolumbar spine. In analogy to initial postoperative in vivo circumstances of human spine motion, the rotational stability was tested in a six degree of freedom spine simulator.

2. Methods

2.1. Specimens

Six fresh frozen human thoracolumbar spines (Th11-L3) were used for stability testing. Specimens were harvested from four male and two female cadavers with a mean age at death of 64 (SD20.3) years and an mean weight of 71.7 (SD8.2) kg. To exclude spinal specimens with possible structural disorders, abnormalities or previous spinal surgery a preoperative computertomography (CT) scan (GE Lightspeed 16, GE Medical Systems, Waukesha, WI, USA) with an European Forearm Phantom (EFP; QRM GmbH, Möhrendorf, Germany) was carried out. This allowed the determination of the cancellous bone mineral density (BMD) for the vertebrae implanted with posterior pedicle fixation. Mean bone mineral density of all vertebras was 87.9 (SD 20.8) mg/cm^3. BMD values in detail are shown in Table 1.

For pre-testing storage, specimens were vacuum sealed in double plastic bags and kept frozen at −30 °C. Specimens were thawed overnight at 6 °C and prepared at room temperature prior to testing. Soft tissue layers were removed, leaving supporting ligaments, capsules and neural structures (e.g. myelon, nerval roots, dural sack) intact. All experiments and procedures were conducted at room temperature, while the specimens were kept moist with isotonic saline solution for the duration of the testing. This procedure was in accordance with different previous publications showing no change in mechanical properties of

Table 1
BMD (bone mineral density in mg/cm^3) of the vertebral bodies Th11-L3 and average BMD of the used specimen numbers 1–6

S. no.	BMD TH11	BMD TH12	BMD L1	BMD L2	BMD L3	av BMD
1	140.7	113.7	106.4	103.5	115.9	116.0
2	101.5	89.1	83.3	69.9	73.5	83.5
3	58.6	54.1	71.9	46.7	40.5	54.4
4	93.1	87.3	73.7	79.9	73.1	81.4
5	105.6	119.2	93.0	88.6	99.5	101.2
6	104.5	93.9	88.9	84.6	83.3	91.0

specimens using this treatment (Panjabi et al., 1985; Wilke et al., 1998a,b).

After isolation and preparation of the tested segments the cranial (Th11) and caudal (L3) ends of the specimens were centered into equal customized metal cups. Ensuring that the middle vertebra (L1) was aligned horizontally (Fig. 1a) both ends of the specimen were then embedded in polymethylmethacrylate (PMMA) cement (Technovit 3040, Heraeus Kulzer, Wehrheim, Germany). To allow segmental movement embedding did not incorporate the articulating parts. Flanges were mounted to the cranial and caudal PMMA blocks of the specimens. The flanges allowed a rigid fixation of the specimens in the spine simulator (Fig. 1b).

Standardized lateral and anterior-posterior X-rays were taken prior to testing of each implantation setting. X-rays were analyzed for segmental alignment changes using angulations between the endplates adjacent to the resected vertebra.

2.2. Biomechanical testing

The biomechanical testing of the spines was carried out according to the recommendations for testing of spinal implants (Wilke et al., 1998a,b). All tests were conducted in a six degree of freedom spine simulator (Fig. 1b). The spine simulator was constructed in accordance to the studies by Knop et al. (2000) and has been modified with additional control and measurement features. The flexibility tests were performed using pure moments of 7.5 Nm in the main motion planes (flexion/extension (My), lateral bending left/right (Mx) and axial rotation left/right (Mz)). The loads induced at the cranial end of the specimen were recorded continuously using a six-component load cell (FT Delta SI 660-60, Schunk Lauffen, Germany). The motion of the bridged segment (L1) was measured using an ultrasound based three-dimensional motion analysis system (Winbiomechanics, Zebris, Isny, Germany) mounted to the adjacent vertebrae Th12 and L2 (Fig. 1a and b).

The recorded data of the six-component load cell and the motion analysis system were used to determine the range of motion (RoM) of the bridged segment Th12-L2. To minimize the viscoelastic effect the specimens were preconditioned with two load cycles and only the third load cycle was used for data evaluation.

2.3. Experimental protocol

First the intact specimens (T11-L3) were loaded in the spine tester in the three main motion planes with pure moments of 7.5 Nm.

After the intact test of the specimens an en-bloc spondylectomy of the first lumbar vertebra was performed followed by a stabilization in accordance to the different implantation settings. After the implantation procedures standardized anterior–posterior and lateral X-rays with the specimen fixed in a customized X-ray jig were taken to verify the correct positioning of the implants.

2.4. En-bloc spondylectomy

All resections and implantations were performed by orthopaedic surgeons trained in the procedure. For

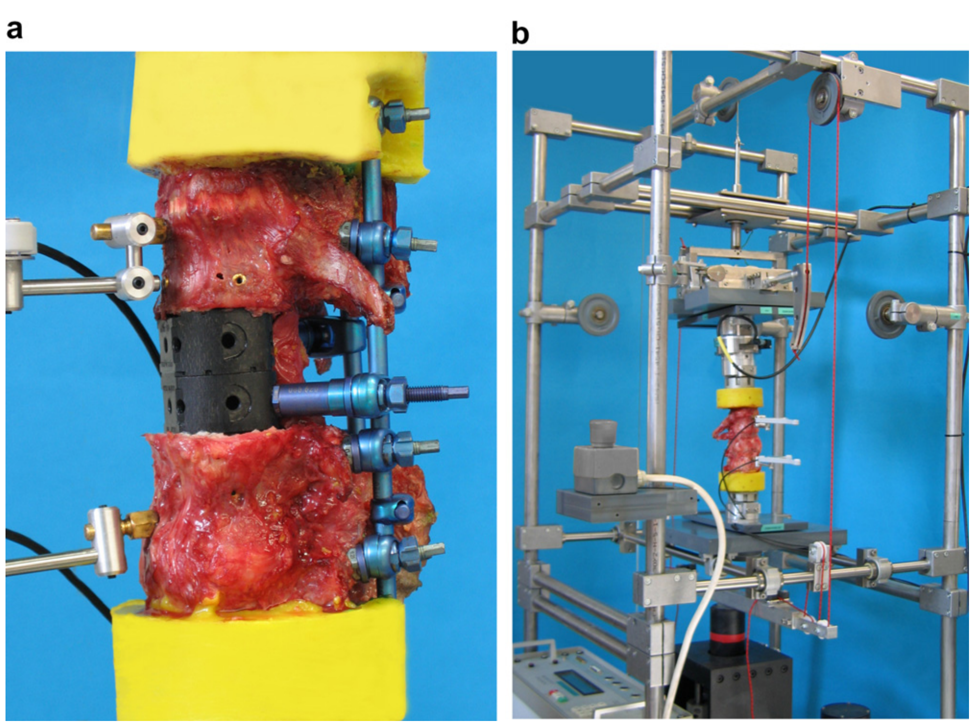

Fig. 1. (a) PMMA embedded specimen with reconstruction of the first lumbar vertebra. (b) Specimen fixed to the six degree of freedom spine simulator.

resection and implantation specimens were fixed in a customized X-ray jig, which allowed the application of an axial preload by dead weights. The simulated defect situation included a mono-segmental vertebrectomy (total en-bloc spondylectomy and additional resection of the adjacent intervertebral discs) of the middle vertebra (L1). The approach and technique were carried out in accordance to previous publications (Disch et al., 2007; Tomita et al., 1997) simulating a solitary posterior approach. En-bloc spondylectomy of L1 was conducted with routinely used clinical instruments. Resections were orientated in accordance to Weinsteins anatomic zone classification of spinal tumour lesions. The spondylectomy includes the zones I–IV (Weinstein, 1989). First, the zones I and II were resected and the dural sack was prepared and mobilized. After intermediate stabilization and cutting of the two adjacent discs zones III–IV were resected en-bloc.

2.5. Implantation

Resection of L1 was followed by reconstruction of the load bearing anterior column with a long fibre carbon composite VBR (Trabis in ostaPek, coLigne AG, Zurich, Switzerland) which was connected to a titanium-made posterior pedicle screw and rod system (eVos, coLigne AG, Zurich, Switzerland) by two artificial pedicles. The artificial pedicles are built up of a threaded rod (Ø 5 mm), which is screwed in the VBR. The rod is the guidance for a sleeve (Ø 10 mm) which has a spherical end that allows a poly-axial fixation of the artificial pedicle to the rod of the internal fixator. The used internal fixator has a poly-axial connection to the pedicle screws with a rod diameter of 6 mm. For standardization 6.25×40 mm pedicle screws were used in all specimens. The cross-section of the VBR was 30×40 mm for all specimens. The height of the VBR was measured of the preoperative CT of the intact specimen and reconstructed with modular VBR parts.

The modular VBR system was assembled without using angulation options. Two titianium transverse connectors with a rod cross-section of 4×3 mm (eVos, coLigne AG, Zurich, Switzerland) were added orthogonal orientated to the posterior fixation system cranially and caudally to the artificial pedicles. Similar to the intraoperative conditions, screw placement was performed under fluroscopy. For implantation of the VBR an axial preload was applied in a specifically designed jig by dead weights. The magnitude of the preload was either 10 N or 100 N according to the implantation setting.

Testing sequences:

1. Intact specimen (IN).
2. VBR and a posterior fixation of 2 upper and 2 lower adjacent segments (T11/Th12; L2/L3), two transverse connectors and an axial compression force of 100 N during implantation (LF).
3. VBR and a posterior fixation of 2 upper and 2 lower adjacent segments (T11/Th12; L2/L3), two transverse connectors and an axial compression force of 10 N during implantation (LN).
4. VBR and a posterior fixation of 1 upper and 1 lower adjacent segment (Th12; L2), two transverse connectors and an axial compression force of 100 N during implantation (SF).
5. VBR and a posterior fixation of 1 upper and 1 lower adjacent segment (Th12; L2), two transverse connectors and an axial compression force of 10 N during implantation (SN).

Test settings 2–5 were carried out in alternating sequences in order to minimize the influence of the testing order.

2.6. Statistics

Statistical analysis was carried out using the SPSS software package (Microsoft Windows release 12.0; SPSS Inc. Chicago, IL, USA). The means and standard deviations for each test sequence were determined. For analysis of differences, the nonparametric paired Wilcoxon test was chosen. Connections between the parameters recorded and the results measured were determined by linear regression analysis. The statistical evaluation was explorative and not adjusted for multiple comparison (Bonferroni test). Therefore the term significant ($P < 0.05$) when used in this study is to be considered a trend.

3. Results

3.1. Biomechanical testing

For the three tested motion planes the measured range of motion (RoM) in axial rotation and lateral bending for the left and right direction was symmetrical and is reported as total RoM (Figs. 2a and 2b). In flexion/extension the RoM was not symmetrical and is split up in the RoM in flexion and in extension (Fig. 2c). All reported val-

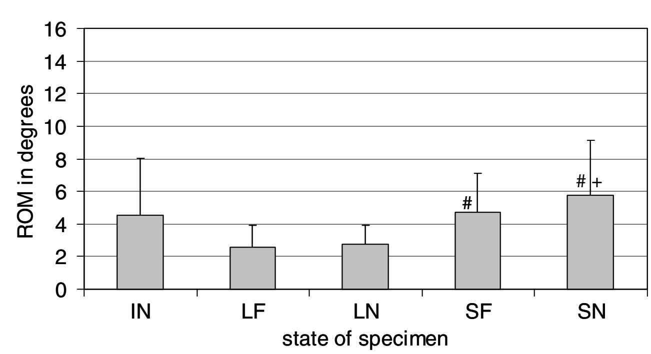

Fig. 2a. Mean values of the range of motion (RoM) for axial rotation, standard deviations (SD) and significances are illustrated; # $P < 0.05$ vs. LF, + $P < 0.05$ vs. LN (IN = intact; LF = long posterior fixation and 100 N compression; LN = long posterior fixation and 10 N compression; SN = short posterior fixation and 10 N compression; SF = short posterior fixation and 100 N compression).

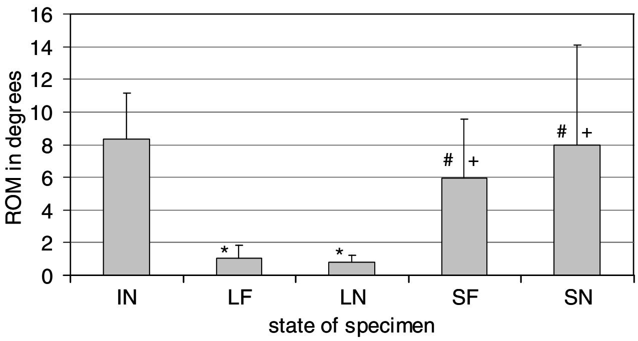

Fig. 2b. Mean values of the range of motion (RoM) for lateral bending, standard deviations (SD) and significances are illustrated; * $P < 0.05$ vs. IN, # $P < 0.05$ vs. LF, + $P < 0.05$ vs. LN (IN = intact; LF = long posterior fixation and 100 N compression; LN = long posterior fixation and 10 N compression; SN = short posterior fixation and 10 N compression; SF = short posterior fixation and 100 N compression).

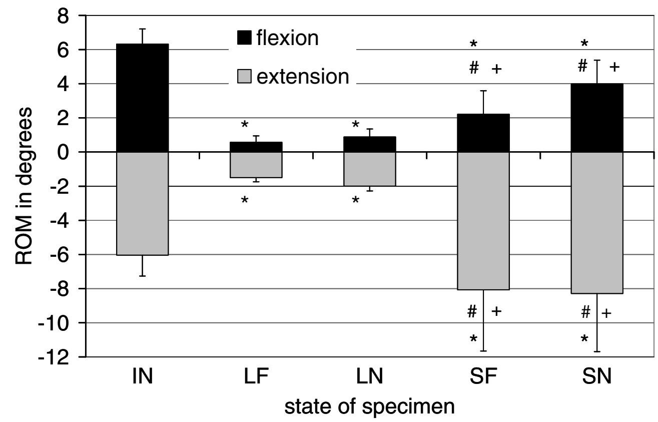

Fig. 2c. Mean values of the range of motion (RoM) for flexion/extension, standard deviations (SD) and significances are illustrated; * $P < 0.05$ vs. IN, # $P < 0.05$ vs. LF, + $P < 0.05$ vs. LN (IN = intact; LF = long posterior fixation and 100 N compression; LN = long posterior fixation and 10 N compression; SN = short posterior fixation and 10 N compression; SF = short posterior fixation and 100 N compression).

ues are the mean and standard deviation of the six tested specimens (Table 2).

3.1.1. Axial rotation

Compared to the intact specimen the short posterior fixation did not stabilize the segment. In contrast to the short posterior fixation the long posterior fixation stabilized the segment. However, none of these differences compared to the intact state was significant. The long posterior fixation was significantly stiffer than the short posterior fixation (Fig. 2a).

3.1.2. Lateral bending

Stabilization of the defect with a long posterior fixation with both axial compression forces significantly decreased the RoM ($P < 0.05$) compared to the intact state and the short posterior fixation. The short posterior fixation with (10 N and 100 N) reduced the RoM compared to the intact state, however, none of these differences were statistically significant. For long posterior fixations, a higher compression force did not change the RoM, while in contrast the RoM in short posterior fixations was slightly reduced (Fig. 2b).

3.1.3. Flexion/extension

Reconstructions using long or short posterior fixations with both axial compression forces significantly decreased the RoM in flexion compared to the intact specimen (Fig. 2c). Long posterior fixation was significantly more stable than short posterior fixation independent of the axial compression force. In extension, the long posterior fixation with both axial compression forces was significantly stiffer than the intact or the short posterior fixation. In contrast to flexion, the short posterior fixation did not have a stabilizing effect compared to the intact specimen and resulted in a significant increase in RoM.

4. Discussion

This is the first report of a biomechanical testing series with a connected VBR pedicle screw fixation system for an en-bloc spondylectomy model. The implant employed in this study was developed for spinal reconstructions following radical surgical resection techniques at the spine. The design is focused on long term outcome by providing large bone graft volume of the VBR, trabecular orientation of the long VBR carbon fibres and to protect the VBR against dislocation by artificial pedicles. However, first of all a reconstruction system has to stabilize the generated defect that results from different grades of tumours and its subsequent surgical resection (Ebihara et al., 2004). Reconstruction was varied by posterior fixation lengths and axial compression forces during implantation.

Table 2
Means and standard deviations (SD) of the RoM (degrees) in the three plains of motion, displayed for right/left or flexion/extension and overall values

	Axial rotation			Lateral bending			Flexion/extension		
	Right	Left	Overall	Right	Left	Overall	Flexion	Extension	Overall
IN	2.1 (±1.6)	−2.4 (±2.0)	4.5 (±3.6)	4.3 (±1.5)	−4.1 (±1.3)	8.4 (±2.8)	6.3 (±0.9)	−6.0 (±1.2)	12.4 (±2.0)
LF	1.2 (±0.7)	−1.2 (±0.4)	2.5 (±1.3)	0.5 (±0.4)	−0.5 (±0.4)	1.0 (±0.8)	0.5 (±0.3)	−1.5 (±0.2)	2.1 (±0.5)
LN	1.3 (±0.7)	−1.3 (±0.6)	2.7 (±1.2)	0.4 (±0.2)	−0.4 (±0.2)	0.8 (±0.4)	0.9 (±0.4)	−2.0 (±0.3)	2.9 (±0.7)
SF	2.3 (±1.2)	−2.4 (±1.2)	4.7 (±3.2)	2.9 (±1.8)	−3.0 (±1.9)	6.0 (±3.6)	2.5 (±1.0)	−7.7 (±3.1)	10.3 (±3.7)
SN	2.9 (±1.6)	−3.0 (±1.7)	5.8 (±3.3)	4.0 (±3.1)	−4.0 (±3.7)	8.0 (±6.2)	4.0 (±1.4)	−8.3 (±3.4)	12.3 (±4.6)

IN = intact; LF = long posterior fixation and 100 N compression; LN = long posterior fixation and 10 N compression; SN = short posterior fixation and 10 N compression; SF = short posterior fixation and 100 N compression.

Our data indicate that short posterior fixation modes (SN/SF) spanning only one adjacent segment may not provide adequate initial three-dimensional stiffness following an en-bloc spondylectomy.

We found the highest intersegmental rotation values of all testing sequences in short posterior fixation modes for extensional movements with an increase in RoM of 37% (SN) and 34% (SF) compared to the intact specimen. Additionally, in axial rotation short posterior fixation failed to restore the stiffness of the intact specimen showing an increase in RoM of 27% (SN) and 4% (SF). However, compared to short posterior fixations, long posterior fixations demonstrated significant more stiffness in all three motion planes. Long posterior fixations (LN/LF) were significantly more stable in extension/flexion (23%/17% of the intact state) and in lateral bending (9%/12% of the intact state) compared to the intact specimen. In axial rotation, long posterior fixations proved to be more stable than the intact specimens.

Increasing axial compression during implantation from 10 N to 100 N slightly affected the construct stiffness without statistical significance. The decrease in RoM with a higher axial compression force was more pronounced for the short posterior fixation.

Most of the biomechanical studies dealing with traumatic or tumoral defects of the spine used only corporectomy models (Eichholz et al., 2004; Knoller et al., 2005; Knop et al., 2000). A corporectomy setting, however, typically leaves the dorsal structures of the spinal column intact and therefore simulates a much smaller defect. Thus, it cannot be compared to en-bloc spondylectomies which by definition impose a complete loss of spinal osteoligamentary continuity of the spinal column. Because other measurement values than the RoM have been used to assess stiffness after reconstruction following en-bloc spondylectomy, the comparison with previous studies remains difficult. Regarding the RoM of the intact specimens, our results are in the range of that reported in other studies (Knop et al., 2000; Vahldiek and Panjabi, 1998), which used the same spinal level, number of motion segments and also applied pure bending moments to assess the range of motion.

Comparing our results to other biomechanical en-bloc spondylectomy studies revealed differences regarding the test set up and the way stiffness was assessed.

Oda et al. (1999) loaded the specimens with bending moments of 4 Nm in flexion/extension and lateral bending applied by a hydraulic material testing machine. They assessed the stiffness of the implant constructs by the axial displacement of the bridged segment at the anterior side using an extensometer fixed over the created defect. For stabilization of the defect they implanted a harms mesh cage as VBR in combination with a long and short posterior fixation. For lateral bending and flexion/extension of the long posterior fixation they also found a significantly higher stiffness relative to both, the short posterior fixation and the intact state. However, in contrast to our study, in lateral bending they showed a significant decrease in stiffness for the short fixation. This might possibly be due to differences in measurement variables. Whereas the angular rotation was used to assess the stiffness in our study, Oda et al. (1999) analyzed the axial displacement in cranio-caudal direction.

Shannon et al. (2004) also used a hydraulic material testing machine to apply bending moments of 4 Nm to spinal specimens with a stabilized en-bloc spondylectomy. They also measured the axial displacement across the bridged defect using an extensometer. Their investigated stabilization methods included a long posterior fixation with no VBR, with an anterior Z-plate supported by a rib graft and an anterior cement and pin construct. They reported the stiffness of the constructs in Nm/degrees. Compared to the intact specimens, following en-bloc spondylectomy they reported higher stiffness values for all implanted reconstructions in all motion planes. This is in accordance with our findings, as all long posterior fixations showed a higher stiffness than the intact specimen.

In spinal stabilization involving VBR combined with posterior fixation the length of the posterior fixation is a major determinant for rigidity of the construct (Eichholz et al., 2004; Vahldiek and Panjabi, 1998). The functional relationship between length of posterior fixation and spinal stiffness following en-bloc spondylectomy and subsequent reconstruction is emphasised by the decreased range of motion in the long posterior fixation group (Fig. 3) when compared to short fixation and intact specimen in this study. This interrelation lends further support to the notion that longer posterior fixation is the causative factor for increased stiffness, in particular during extension when there is no anterior support to counteract increasingly developing motion and instability. Compared to a short posterior fixation, with a long posterior fixation the load acting on the spine is transferred across the defect by 8

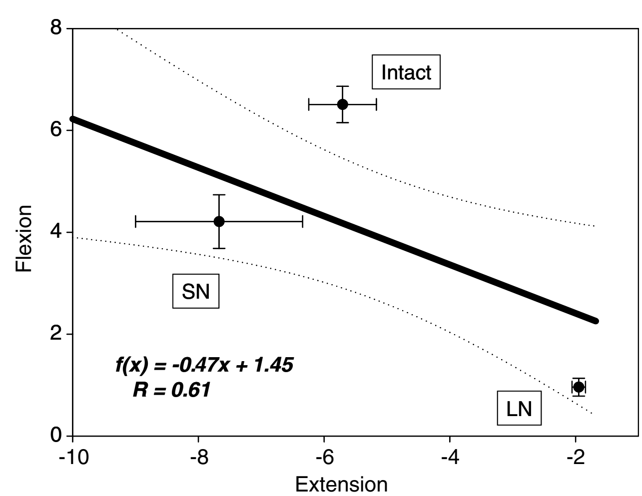

Fig. 3. Regression analysis between values for extension and flexion of the long (LN) and short fixation (SN) without compression and intact state as tested in the six degrees of freedom spine simulator. Values are mean, SEM (standard error of mean), R = regressions coefficient, dotted lines represent 95% confidence intervals.

instead of 4 pedicle screws, thereby reducing the bending moment acting on each screw (Brodke et al., 2001; Oda et al., 1999).

Pflugmacher et al. (2004) investigated the effect of various expandable or non-expandable VBRs on primary stiffness. They concluded, that for a combined anterior-posterior stabilization the type of VBR had only a minor influence on primary stiffness of the treated segment. In en-bloc spondylectomy models it was shown that titanium mesh cages (Oda et al., 1999) as well as anterior PMMA constructs (Shannon et al., 2004) when combined with multilevel posterior fixations were able to provide more stiffness than the intact specimens.

Various authors (Knoller et al., 2005; Oda et al., 1999; Shannon et al., 2004; Vahldiek and Panjabi, 1998) assessed 360° stabilizations for corporectomy and en-bloc spondylectomy models and demonstrated higher primary stiffness values compared to singular posterior or anterior stabilization methods. A 360° stabilization is biomechanically favourable, as it stabilises a spinal segment on both sides of the center of rotation. However, when transferring these results to the clinical application, one must consider that additional anterior fixation for 360° stabilization requires expansion of surgery in an already extensive operation. This may be associated with an increase in surgical risk factors, i.e. enlarged surgical approaches, increased infection risk, more blood loss and prolonged operation time. Therefore, extensive spine tumour surgery experience and an individual decision making depending on patient‘s characteristics appears essential.

Reduced to the bony and ligamentary structures biomechanical testing set ups of the spine have known limitations (Oda et al., 1999; Shannon et al., 2004). Most of all, the influence of the absent muscles on biomechanical characteristics cannot be evaluated. Comparing with other studies or transferring these gathered results to the clinical situation remains difficult. Variable specimen characteristics (age, BMD, spinal level, species), testing conditions and testing sequences are used. In addition, the influence of secondary factors such as tissue healing, e.g. bony ingrowth can not be investigated in ex vivo experiments. Non the less, biomechanical in vitro testing can be used to assess the initial influence of implants on the stiffness of the treated segment. According to the recommendations on standardized spinal implant testing (Wilke et al., 1998a,b) specimens were loaded with pure moments and the range of motion was used to compare the stiffness of the investigated implant settings. Loading specimens with pure moments in the three main motion planes is a widely accepted method for spinal implant testing (Panjabi, 1988; Wilke et al., 1994).

5. Conclusion

Combination of long posterior fixations with the investigated connected VBR pedicle screw system showed superior initial stiffness following an en-bloc spondylectomy when compared to the intact state. Short posterior fixations provided significant lower stiffness even when combined with higher axial compression loads during implantation.

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