The Effect of Polymethyl Methacrylate Augmentation on the Primary Stability of Cannulated Bone Screws in an Anterolateral Plate in Osteoporotic Vertebrae: A Human Cadaver Study

Matthias Rüger[1] Richard M. Sellei[2] Marcus Stoffel[3] Christian von Rüden[1]

[1] Department of Reconstructive Orthopaedics, Trauma Center Murnau, Murnau, Bavaria, Germany
[2] Department of Trauma Surgery, RWTH Aachen University Medical Center, Aachen, Germany
[3] IAM Institute of General Mechanics, RWTH Aachen University, Aachen, Germany

Global Spine J 2016;6:46–52.

Address for correspondence Matthias Rüger, MD, Department of Reconstructive Orthopaedics, Trauma Center Murnau, Professor-Küntscher-Str. 8, 82418 Murnau a. Staffelsee, Germany (e-mail: matthias.rueger@bgu-murnau.de).

Abstract

Study Design Cohort study.

Objective Expandable anterolateral plates facilitate the reduction of posttraumatic deformities of thoracolumbar spine injuries and are commonly used in cases of unstable injuries or compromised bone quality. In this in vitro study, the craniocaudal yield load of the osseous fixation of an anterior angular stable plate fixation system and the effect of polymethyl methacrylate (PMMA) screw augmentation on the primary stability of the screw–bone interface during kyphosis reduction was evaluated in 12 osteoporotic human thoracolumbar vertebrae.

Methods The anterolateral stabilization device used for this study is comprised of two swiveling flanges and an expandable midsection. It facilitates the controlled reduction of kyphotic deformities in situ with a geared distractor. Single flanges were attached to 12 thoracolumbar vertebrae. Six specimens were augmented with PMMA by means of cannulated bone screws. The constructs were subjected to static, displacement-controlled craniocaudal loading to failure in a servohydraulic testing machine.

Results The uncemented screws cut out at a mean 393 ± 66 N, whereas the cemented screws showed significantly higher yield load of 966 ± 166 N ($p < 0.02$). We detected no significant correlation between bone mineral density and yield load in this setting.

Conclusion Our results indicate that PMMA augmentation is an effective method to increase two- to threefold the primary stability of the screw–bone interface of an anterolateral spine stabilization system in osteoporotic bone. We recommend it in cases of severely compromised bone quality to reduce the risk of screw loosening during initial kyphosis correction and to increase long-term construct stability.

Keywords
- thoracolumbar spine injury
- osteoporosis
- PMMA augmentation
- yield load
- anterolateral stabilization

received
January 8, 2015
accepted after revision
April 23, 2015
published online
June 15, 2015

DOI http://dx.doi.org/10.1055/s-0035-1555659.
ISSN 2192-5682.

Introduction

Anterior stabilization secondary to posterior instrumentation is an established treatment strategy for unstable thoracolumbar spinal injuries and typically includes spondylodesis with a bone graft, intervertebral cage, or vertebral body replacement.[1–3] Anterolateral plating systems are utilized to enhance the overall construct stability and reduce the risk of cage subsidence in cases of highly comminuted or multilevel burst fractures, vertebrectomy, or severely compromised bone quality.[4–8] More recent designs are expandable to adapt to individual anatomic situations and to facilitate the reduction of sagittal plane deformities from anterior.[9–11] Kyphosis reduction via pedicle screws remains the gold standard, as it is a reliable and effective technique that benefits from strong pedicular bone stock and a relatively large cross-sectional bone contact area of the long transpedicular screw trajectory. In contrast, bone screws of anterolateral devices are shorter than pedicle screws and are typically positioned monocortically to avoid vascular damage. As a result, anterolateral bone screws have less cortical bone support and less cross-sectional cancellous bone contact area for load distribution than pedicle screws of the same segment, which raises the question if the anterolateral screw fixation bears an increased risk for craniocaudal cutting out during kyphosis reduction, particularly in osteoporotic bone. For comparison, we studied the effect of polymethyl methacrylate (PMMA) screw augmentation on the primary interface strength.

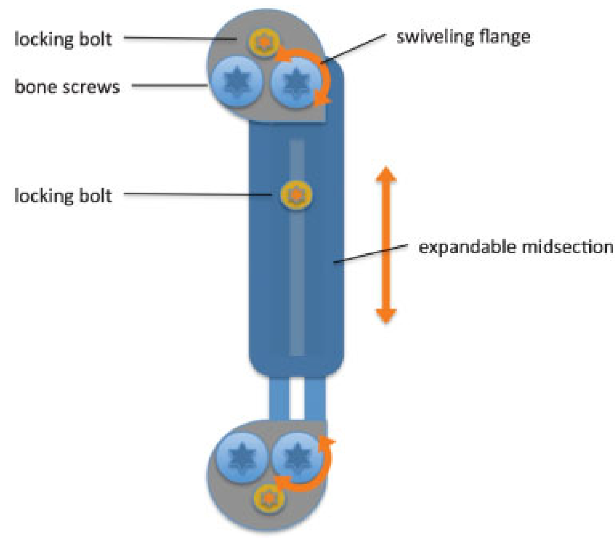

Fig. 1 Schematic of anterolateral device used in this study. Two bone screws affix a swiveling flange laterally to a vertebral body, with the posterior screw being the center of rotation. A locking bolt locks this joint. The midsection is expandable by means of a geared distractor that attaches to the flanges. Once the desired sagittal plane correction is achieved, the midsection can be locked as well.

Materials and Methods

The anterolateral plating system used in this study is comprised of two swiveling flanges and a telescopic midsection (►**Fig. 1**; DePuy Synthes, Oberndorf, Switzerland). It facilitates in situ deformity correction by means of a geared distractor that expands the midsection. Each flange mounts to the lateral face of a vertebral body with two monocortical, angular-stable bone screws. The screws have a conical shape and converge at an angle of 5 degrees. The anterior bone screws are cannulated for optional PMMA augmentation. Three Torx screws lock the swiveling joints and telescopic midsection in their final position.

Preparation

Twelve fresh-frozen thoracolumbar vertebrae from two human donors with reduced bone mineral density (BMD) were prepared. All soft tissues including the intervertebral disks were dissected according to standard,[13,14] leaving only the osseous structures intact. The BMD of each vertebral body was measured with computed tomography using a standard phantom provided by the manufacturer (Siemens, Erlangen, Germany). The properties of the anatomic specimens are summarized in ►**Table 1**. Under fluoroscopic control (BV25, Philips, Eindhoven, The Netherlands), the flanges were mounted to the vertebral bodies with bone screws of equal length (30 mm) to maintain a constant cross-sectional bone contact area. The instrumented vertebrae selected for screw augmentation received 4 mL of high-viscosity PMMA cement (Vertecem, DePuy Synthes, Oberndorf, Switzerland) into the anterior screw according to specifications. The PMMA was prepared using an electronic viscometer to determine the ideal interval for cement injection.

Biomechanical Testing

The biomechanical trials were conducted in a servohydraulic testing machine (Flextest II M, MTS, Eden Prairie, Minnesota, United States), fitted with a 10-kN load cell (operating at 0.5% error). The specimens were held in place by two metal clamps. The actuator of the servohydraulic testing machine exerted a craniocaudal force, thereby simulating kyphosis correction. The mechanical testing was conducted statically and was displacement-controlled to detect yield load and failure mode and to compensate for any viscoelastic phenomena. The displacement rate was set to 0.02 mm/s, and the abort criteria were set to 6-mm displacement or 1,500 N. After reaching the abort criteria, the servohydraulic testing machine reduces displacement at a rate of 0.02 mm/s, until the recorded load is 0. The remaining displacement then reflects the extent of plastic deformation. All data was recorded at 20 Hz. This setup was chosen according to standard protocols and is given in ►**Fig. 2**.[12–14]

Data Analysis

Statistical analysis was performed using SPSS 16 (SPSS Inc., Chicago, Illinois, United States) for Apple OS X. For inferential statistics, we applied the Mann-Whitney *U* test and calculated Pearson correlation coefficients for independent samples, when applicable.

Table 1 Summary of threshold loads marking begin of plastic deformation, and respective displacements

	Donor 1			Donor 2		
Fixation	No.	Level	BMD (mg HA/cm^3)	No.	Level	BMD (mg HA/cm^3)
Cemented	P1	T12	63.4	P4	T12	98.4
	P2	L2	66.9	P5	L1	106
	P3	T11	72.6	P6	T11	152.7
Uncemented	P7	L1	65.5	P10	L2	103.5
	P8	L3	67.5	P11	L3	112.4
	P9	L4	84.1	P12	L4	158.3
Mean			70.0			121.9

Abbreviations: BMD, bone mineral density; HA, hydroxyapatite.
Note: Donor 1 was an 80-year-old woman; donor 2 was a 55-year-old man. P1–P6 were augmented, P7–12 were nonaugmented.

Results

The Influence of Cement Augmentation on Screw Anchorage

The cemented screws demonstrated a significantly stronger resistance against axial loading and exhibited a failure mode resembling linear-elastic behavior in the ascending branch of the graph (►**Fig. 3**). The uncemented screws showed an almost linear failure mode. Cutting out started early and proceeded linearly (►**Fig. 4**). ►**Fig. 5** gives a summary of the different failure modes and end points of each run. The cemented screws (P1 to P6, dashed lines) clearly cut out less than the noncemented samples. ►**Fig. 6** shows two specimens from each group, illustrating the higher bearing resistance of the PMMA-cemented screws in the anterior screw canal. Computed tomography scans revealed an oval anterior screw canal with a constant height-to-width ratio along its entire length (►**Fig. 7**).

The yield loads of the uncemented specimens ranged from 289 to 451 N, and the corresponding displacements were 0.95 to 1.35 mm. In contrast, the cemented specimens exhibited yield loads from 735 to 1,131 N and displacements from 1.64 to 2.1 mm (►**Table 1**). All the cemented samples reached the load limit of 1,500 N at an average displacement of 3.9 mm, and all the uncemented samples reached the maximum displacement of 6 mm at an average force of 1,105 N. The cemented and uncemented screws exhibited significantly different failure loads ($p < 0.05$) and displacements at failure ($p < 0.05$).

Influence of Bone Quality on Screw Purchase

The mean BMD of the specimens from donor 1 was 70.0 ± 7.6 mg hydroxyapatite (HA)/cm^3, qualifying as highly osteoporotic, and the BMD of the specimens from donor 2 was 121.9 ± 26.5 mg HA/cm^3, which qualifies as osteopenic (►**Table 2**).

All the specimens exhibited a significantly different ($p < 0.02$) failure mode and yield load with cement augmentation (960.3 ± 203.6 N) and without cement augmentation (410.7 ± 64.7 N). The displacement at failure was 1.15 ± 0.2 mm in the uncemented group versus 1.91 ± 0.24 mm in the cemented group ($p < 0.02$). Likewise, the osteopenic specimens demonstrated yield loads of 377.0 ± 77.2 N in the uncemented group versus 974.5 ± 6.36 N in the cemented group ($p < 0.002$), and displacement at failure was 1.13 ± 0.125 mm in the uncemented group versus 1.73 ± 0.03 mm in the cemented group ($p < 0.01$). We did not detect a correlation between bone

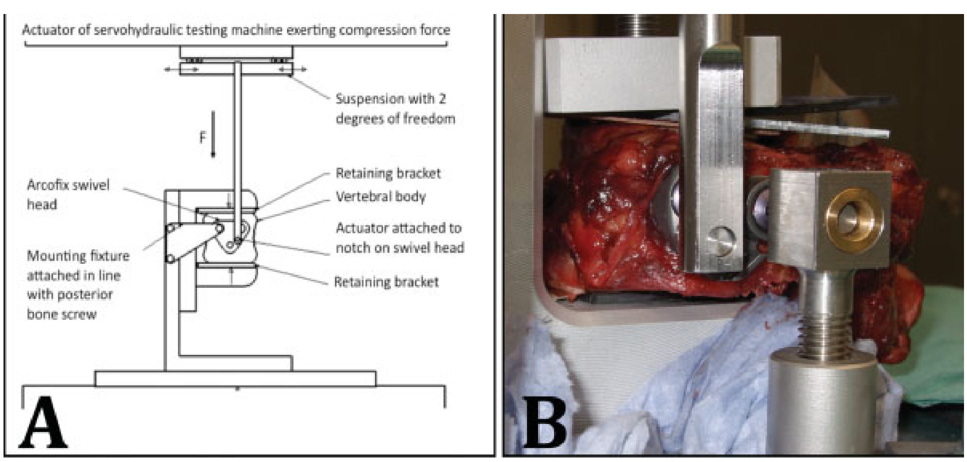

Fig. 2 Biomechanical test setup. The posterior screw is fixed to the loading jig of a servohydraulic testing machine, and the anterior screw is loaded by the actuator, simulating the compressive loading during intraoperative kyphosis correction. (A) Diagram and (B) corresponding photograph.

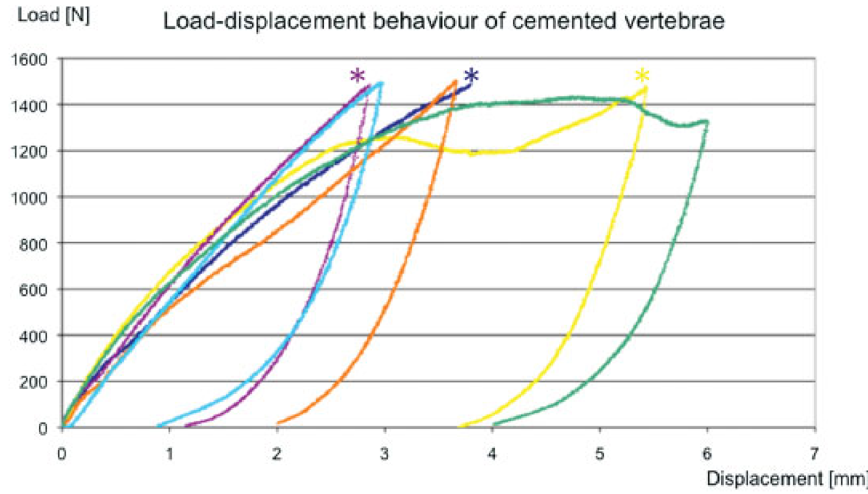

Fig. 3 Load-displacement plots of augmented specimens. *Specimens with the lowest bone mineral density in the group (P1 to P3).

mass and either yield load ($R^2 = 0.015$) or displacement ($R^2 = 0.005$) at failure. The relative effect of the PMMA augmentation on the bearing resistance was again not significantly different between the osteopenic bone and osteoporotic bone. The average increase in the yield load was 597.5 (+258.5%) N in the osteoporotic bone and 549.7 (+234%) N in the osteopenic bone. The average increase of displacement at failure was 0.6 mm in the osteoporotic bone and 0.77 mm in the osteopenic bone.

Discussion

Rationale

The anterolateral device used in this study enables the surgeon to correct sagittal plane deformities by means of a geared distractor that attaches directly to the swiveling flanges on either end of a telescopic midsection. It can be applied after posterior instrumentation or in an anterior-only procedure in combination with an intervertebral cage, vertebral body replacement device, or bone graft.[15–17] We hypothesized that the limited length and monocortical fixation of the anterolateral bone screws could provide insufficient purchase in osteoporotic bone, leading to craniocaudal cutout of the bone screws during kyphosis correction and resulting in early construct failure. Although the yield load of pedicle screws had been thoroughly investigated prior to our experiment, we found no such data for anterolateral bone screws. We chose two human specimens with reduced bone mass as our main focus was to study screw purchase and the relative effect of PMMA augmentation in this patient group. The bone mass between both donors was significantly different, so any effect of BMD on yield load would be detected as well.

Study Design

To elucidate the craniocaudal yield load of the bone–screw interface during initial reduction, we chose a static displacement-controlled routine over a cyclic testing sequence. The resulting plots represent hysteresis curves. On these graphs,

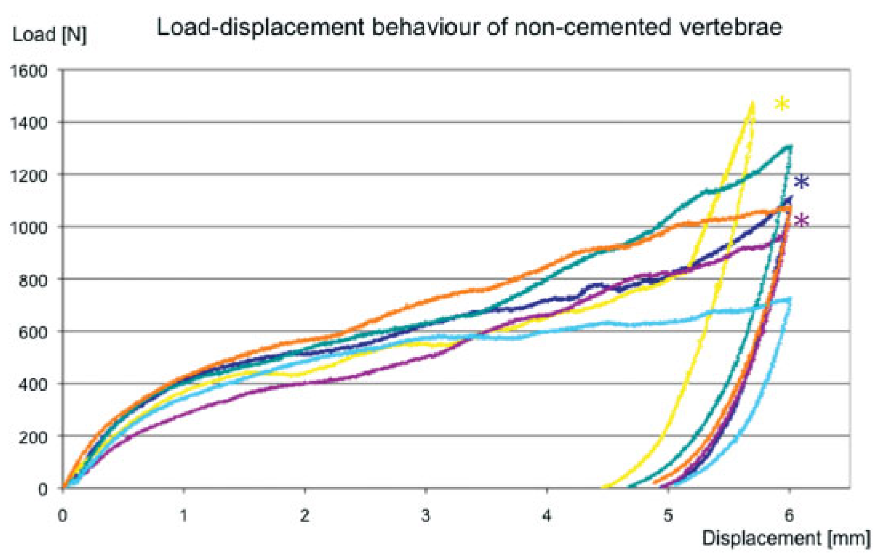

Fig. 4 Load-displacement plots of nonaugmented specimens. *Specimens with the lowest bone mineral density (P7 to P9).

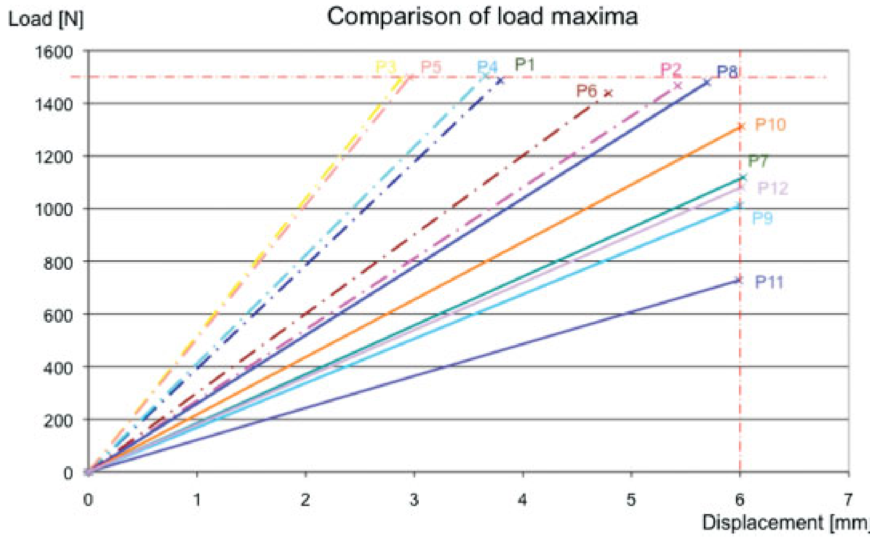

Fig. 5 Graphical representation of different failure modes translating to different end points (maximum load or displacement). P1 to P6 (dashed lines) are augmented; P7 to P12 are nonaugmented samples.

the first inflection point marks the beginning of screw cutout and coincides with the yield load. By interpreting the relaxation of bone as a strictly elastic-plastic process, the difference between maximal displacement and hysteresis approximates the yield load of plastic deformation.

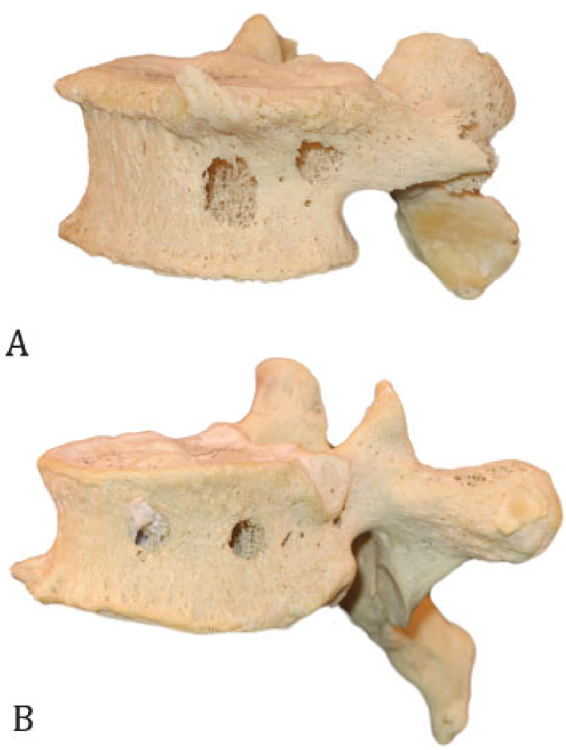

Fig. 6 Macerated vertebral bodies after completion of testing. Note the circular orifice with traces of polymethyl methacrylate in an augmented specimen (B) versus oval orifice of the anterior screw canal in a nonaugmented specimen (A).

The Influence of Cement Augmentation on Screw Anchorage

The use of PMMA augmentation led to a distinctly different linear-elastic failure mode under craniocaudal loading. Gross inspection of the macerated specimens and subsequent computed tomography demonstrated collateral cutting out of the uncemented screws, resulting in a complete loss of osseous fixation. Both the different failure modes and the different yield loads indicate that screw augmentation significantly enhanced bearing resistance under these specific conditions.

The Influence of Bone Quality on Screw Anchorage

The mean yield load for the uncemented specimens in our experiment is consistent with recent data published by Baluch et al,[18] who reported a craniocaudal toggle load to failure of 300 N for noncortical pedicle screws and 398 N for screws with monocortical transpedicular fixation. In our experiment, we were unable to establish a correlation between BMD and yield load. Trabecular bone comprises a network of both craniocaudally oriented rods and smaller interconnecting transversal struts. Osteoporotic bone is characterized by a rarefied trabecular structure, predominant in the cancellous regions. Screw purchase in bone relies on a dense network of trabeculae to interlock with the threads. Several publications demonstrated the correlation of BMD and anchorage of spinal implants. Reinhold, Goldhahn, and other authors reported that the pullout resistance of vertebral screws largely depends on cancellous bone quality.[19–24] Another study found that the pullout resistance of pedicle screws is largely dependent on BMD if the cortex is thinner than 2 mm.[25] However, this data applies to axial pullout forces only. It could be argued that the craniocaudal bearing resistance in the human vertebra depends not only on cancellous mineral mass (as measured by dual-energy X-ray absorptiometry) but on cortical thickness as well.[26] Our result is partly corroborated by Disch et al,[27] who tested the rigidity

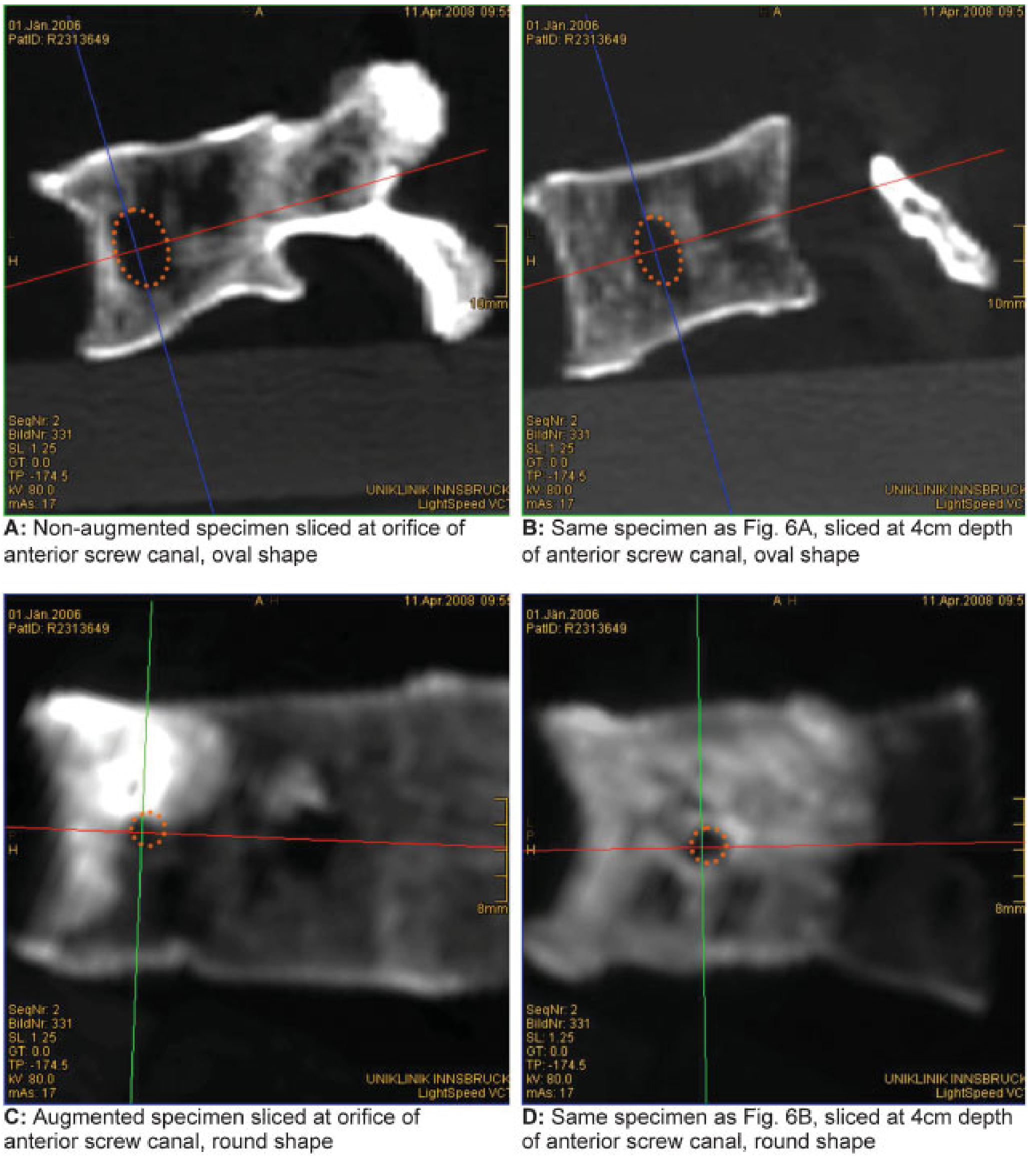

Fig. 7 Computed tomography scans of the specimens shown in ►**Figs. 6A** and **6B**. Sagittal slices exactly perpendicular to anterior screw canal at orifice (A, C) and at 4-cm depth (B, D). (A, B) Oval shape of screw canal with similar height-to-width ratio throughout entire screw canal indicates cutting out while maintaining an effective angular stable connection between swivel head and screw. (C, D) polymethyl methacrylate augmentation drastically reduces cutting out. Screw canal remains circular throughout its entire depth.

Table 2 Summary of human anatomic specimens used in this study, including BMD as measured with qCT

	Osteoporotic bone (70.0 mg/cm^3)				Osteopenic bone (121.9 mg/cm^3)			
	No.	Level	Displ. (mm)	Load (N)	No.	Level	Displ. (mm)	Load (N)
Augmented	P1	T12	1.68	982	P4	T12	1.64	735
	P2	L2	1.75	970	P5	L1	2.10	1131
	P3	T11	1.71	979	P6	T11	2.00	1,015
Nonaugmented	P7	L1	1.09	433	P10	L2	1.35	445
	P8	L3	1.27	409	P11	L3	0.95	336
	P9	L4	1.03	289	P12	L4	1.14	451

Abbreviations: BMD, bone mineral density; Displ, displacement; qCT, quantitative computed tomography.
Note: The female specimens qualify for osteoporosis, the male specimens represent osteopenic bone

of the implant used in our study against a nonlocking plate design in an in vitro corpectomy model and demonstrated a poor correlation of BMD and yield load for the locking design ($R^2 = 0.12$), whereas the nonlocking plate correlated strongly with BMD ($R^2 = 0.90$).

Conclusion

PMMA augmentation improves the bearing resistance of an anterolateral plating system during in situ kyphosis reduction by two- to threefold. We found no correlation between bone mass and the bearing resistance of the bone screws under static craniocaudal loading, nor between BMD and the effectiveness of cement augmentation. After augmentation, despite their significantly different BMD, all specimens returned similar results in terms of both yield load and displacement at failure. We conclude that in cases of poor bone quality, cement augmentation of the anterolateral bone screws is an effective method to improve primary screw purchase, overall construct rigidity, and long-term survival and may also reduce the risk of cage subsidence.

Disclosures
Matthias Rüger, none
Richard M. Sellei, none
Marcus Stoffel, none
Christian von Rüden, none

References

1 Shono Y, McAfee PC, Cunningham BW. Experimental study of thoracolumbar burst fractures. A radiographic and biomechanical analysis of anterior and posterior instrumentation systems. Spine (Phila Pa 1976) 1994;19(15):1711–1722
2 James KS, Wenger KH, Schlegel JD, Dunn HK. Biomechanical evaluation of the stability of thoracolumbar burst fractures. Spine (Phila Pa 1976) 1994;19(15):1731–1740
3 Pflugmacher R, Schleicher P, Schaefer J, et al. Biomechanical comparison of expandable cages for vertebral body replacement in the thoracolumbar spine. Spine (Phila Pa 1976) 2004;29(13): 1413–1419
4 Dick JC, Brodke DS, Zdeblick TA, Bartel BD, Kunz DN, Rapoff AJ. Anterior instrumentation of the thoracolumbar spine. A biomechanical comparison. Spine (Phila Pa 1976) 1997;22(7):744–750
5 Lange U, Edeling S, Knop C, et al. Anterior vertebral body replacement with a titanium implant of adjustable height: a prospective clinical study. Eur Spine J 2007;16(2):161–172
6 Thalgott JS, Kabins MB, Timlin M, Fritts K, Giuffre JM. Four year experience with the AO Anterior Thoracolumbar Locking Plate. Spinal Cord 1997;35(5):286–291
7 Breeze SW, Doherty BJ, Noble PS, LeBlanc A, Heggeness MH. A biomechanical study of anterior thoracolumbar screw fixation. Spine (Phila Pa 1976) 1998;23(17):1829–1831
8 Brodke DS, Gollogly S, Bachus KN, Alexander Mohr R, Nguyen BK. Anterior thoracolumbar instrumentation: stiffness and load sharing characteristics of plate and rod systems. Spine (Phila Pa 1976) 2003;28(16):1794–1801
9 Ogon M, Haid C, Krismer M, Sterzinger W, Bauer R. Comparison between single-screw and triangulated, double-screw fixation in anterior spine surgery. A biomechanical test. Spine (Phila Pa 1976) 1996;21(23):2728–2734
10 Schreiber U, Bence T, Grupp T, et al. Is a single anterolateral screw-plate fixation sufficient for the treatment of spinal fractures in the thoracolumbar junction? A biomechanical in vitro investigation. Eur Spine J 2005;14(2):197–204
11 Chou D, Larios AE, Chamberlain RH, et al. A biomechanical comparison of three anterior thoracolumbar implants after corpectomy: are two screws better than one? J Neurosurg Spine 2006; 4(3):213–218
12 Wilke HJ, Wenger K, Claes L. Testing criteria for spinal implants: recommendations for the standardization of in vitro stability testing of spinal implants. Eur Spine J 1998;7(2):148–154
13 Ashman RB, Bechtold JE, Edwards WT, Johnston CE II, McAfee PC, Tencer AF. In vitro spinal arthrodesis implant mechanical testing protocols. J Spinal Disord 1989;2(4):274–281
14 Panjabi MM, Krag M, Summers D, Videman T. Biomechanical time-tolerance of fresh cadaveric human spine specimens. J Orthop Res 1985;3(3):292–300
15 Sasso RC, Best NM, Reilly TM, McGuire RA Jr. Anterior-only stabilization of three-column thoracolumbar injuries. J Spinal Disord Tech 2005;18(Suppl):S7–S14
16 Wood KB, Bohn D, Mehbod A. Anterior versus posterior treatment of stable thoracolumbar burst fractures without neurologic deficit: a prospective, randomized study. J Spinal Disord Tech 2005;18 (Suppl):S15–S23
17 Eichholz KM, Hitchon PW, From A, et al. Biomechanical testing of anterior and posterior thoracolumbar instrumentation in the cadaveric spine. Invited submission from the Joint Section Meeting on Disorders of the Spine and Peripheral Nerves, March 2004. J Neurosurg Spine 2004;1(1):116–121
18 Baluch DA, Patel AA, Lullo B, et al. Effect of physiological loads on cortical and traditional pedicle screw fixation. Spine (Phila Pa 1976) 2014;39(22):E1297–E1302
19 Reinhold M, Schwieger K, Goldhahn J, Linke B, Knop C, Blauth M. Influence of screw positioning in a new anterior spine fixator on implant loosening in osteoporotic vertebrae. Spine (Phila Pa 1976) 2006;31(4):406–413
20 Goldhahn J, Seebeck J, Frei R, Frenz B, Antoniadis I, Schneider E. New implant designs for fracture fixation in osteoporotic bone. Osteoporos Int 2005;16(2, Suppl 2):S112–S119
21 Goldhahn J, Reinhold M, Stauber M, et al. Improved anchorage in osteoporotic vertebrae with new implant designs. J Orthop Res 2006;24(5):917–925
22 Lim TH, An HS, Evanich C, Hasanoglu KY, McGrady L, Wilson CR. Strength of anterior vertebral screw fixation in relationship to bone mineral density. J Spinal Disord 1995;8(2):121–125
23 Gilbert SG, Johns PC, Chow DC, Black RC. Relation of vertebral bone screw axial pullout strength to quantitative computed tomographic trabecular bone mineral content. J Spinal Disord 1993; 6(6):513–521
24 Halvorson TL, Kelley LA, Thomas KA, Whitecloud TS III, Cook SD. Effects of bone mineral density on pedicle screw fixation. Spine (Phila Pa 1976) 1994;19(21):2415–2420
25 Seebeck J, Goldhahn J, Städele H, Messmer P, Morlock MM, Schneider E. Effect of cortical thickness and cancellous bone density on the holding strength of internal fixator screws. J Orthop Res 2004;22(6):1237–1242
26 Eswaran SK, Gupta A, Adams MF, Keaveny TM. Cortical and trabecular load sharing in the human vertebral body. J Bone Miner Res 2006;21(2):307–314
27 Disch AC, Knop C, Schaser KD, Blauth M, Schmoelz W. Angular stable anterior plating following thoracolumbar corpectomy reveals superior segmental stability compared to conventional polyaxial plate fixation. Spine (Phila Pa 1976) 2008;33(13): 1429–1437