Eur Spine J (1995) 4: 114–122

European Spine Journal

New means in spinal pedicle hook fixation

A biomechanical evaluation

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Summary. Pedicle hooks which are used as an anchorage for posterior spinal instrumentation may be subjected to considerable three-dimensional forces. In order to achieve stronger attachment to the implantation site, hooks using screws for additional fixation have been developed. The failure loads and mechanisms of three such devices have been experimentally determined on human thoracic vertebrae: the Universal Spine System (USS) pedicle hook with one screw, a prototype pedicle hook with two screws and the Cotrel-Dubousset (CD) pedicle hook with screw. The USS hooks use 3.2-mm self-tapping fixation screws which pass into the pedicle, whereas the CD hook is stabilised with a 3-mm set screw pressing against the superior part of the facet joint. A clinically established 5-mm pedicle screw was tested for comparison. A matched pair experimental design was implemented to evauluate these implants in constrained (series I) and rotationally unconstrained (series II) posterior pull-out tests. In the constrained tests the pedicle screw was the strongest implant, with an average pull-out force of 1650 N (SD 623 N). The prototype hook was comparable, with an average failure load of 1530 N (SD 414 N). The average pull-out force of the USS hook with one screw was 910 N (SD 243 N), not significantly different to the CD hook's average failure load of 740 N (SD 189 N). The result of the unconstrained tests were similar, with the prototype hook being the strongest device (average 1617 N, SD 652 N). However, in this series the difference in failure load between the USS hook with one screw and the CD hook was significant. Average failure loads of 792 N (SD 184 N) for the USS hook and 464 N (SD 279 N) for the CD hook were measured. A pedicular fracture in the plane of the fixation screw was the most common failure mode for USS hooks. The hooks usually did not move from their site of implantation, suggesting that they may be well-suited for the so-called segmental spinal correction technique as used in scoliosis surgery. In contrast, the CD hook disengaged by translating caudally from its site of implantation in all cases, suggesting a mechanical instability. The differences in observed hook failure modes may be a function of the type and number of additional fixation screws used. These results suggest that additional screw fixation allows stable attachment of pedicle hooks to their implantation site. Hooks using additional fixation screws passing obliquely into the pedicle apparently provide the most rigid attachment. The second fixation screw of the prototype hook almost doubles the fixation strength. Thus, the prototype hook might be considered as an alternative to the pedicle screw, especially in the upper thoracic region.

Key words: Thoracic spine – Biomechanics – Pedicle screw – Pedicle hook – Pull-out test

Received: 27 April 1994
Revised: 18 July 1994
Accepted: 4 August 1994

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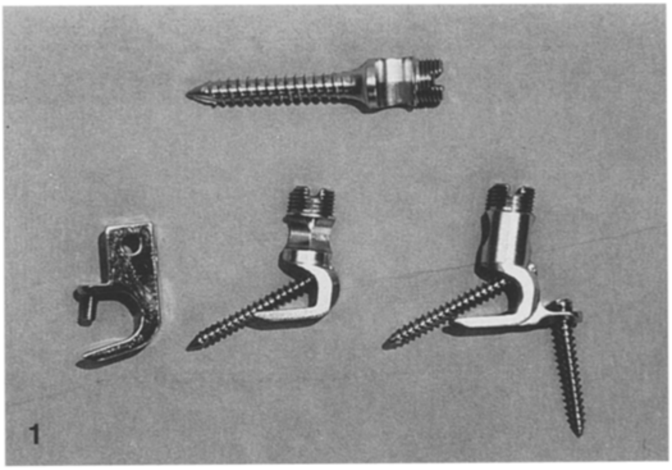
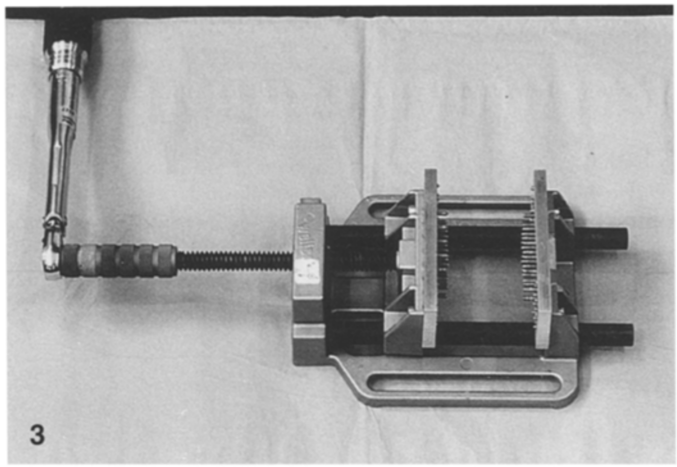
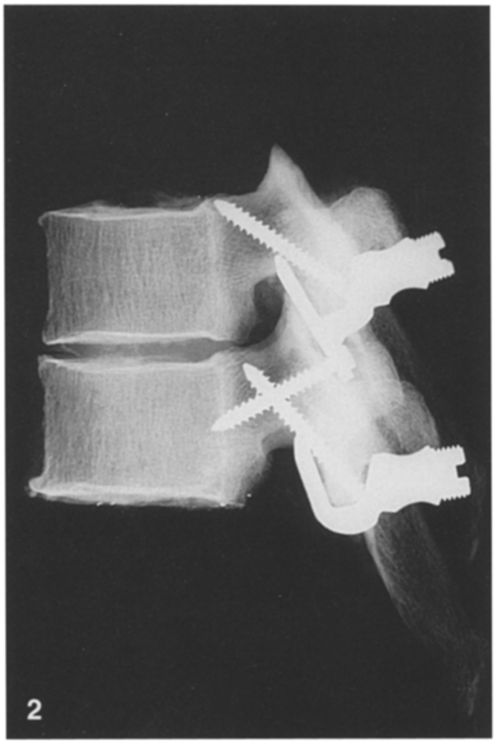
In 1962 Harrington introduced hooks and rods as a means to apply corrective forces in spinal surgery. The hooks were anchored under the vertebral laminae at both ends of the Harrington instrumentation, allowing distraction and compression. Major complications encountered were mechanical failure of the instrumentation and/or disengagement of the hooks, resulting in loss of the original correction [4, 11, 12, 14]. Concentrated hook forces on thin laminae resulted in fixation failure due to fracture of the lamina or bone resorption [8]. Many modifications to the instrumentation and the shape of the hooks addressed these problems.

The Harrington distraction-compression instrumentation basically addresses only the frontal curve abnormality [27]. A negative influence on the sagittal spinal contour can lead to the so-called flatback-syndrome in the lumbar region, causing decompensation in flexion and pain [17]. Recently, systems designed to apply corrective forces to the spinal deformity in all three dimensions have been introduced, e.g. the Cotrel-Dubousset (CD) Instru-







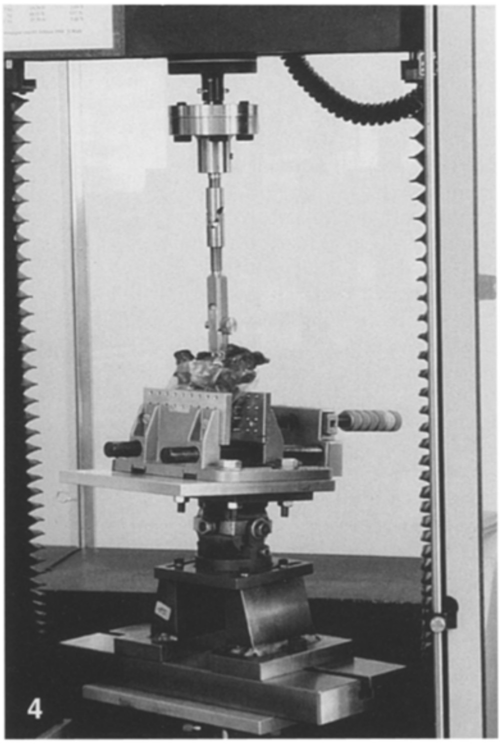


Fig. 1. The four implants tested: (*bottom left*) CD pedicle hook with 3-mm diameter set screw; (*bottom middle*) USS pedicle hook with 3.2-mm-diameter, self-tapping fixation screw; (*bottom right*) prototype pedicle hook with two 3.2-mm-diameter, self-tapping fixation screws; and (*top*) USS 5-mm diameter pedicle screw

Fig. 2. Lateral X-ray of a functional spinal unit (T6–7) with the USS hook implanted inferior and the USS prototype hook implanted superior. The additional fixation screws pass into the pedicles

Fig. 3. The mechanical clamp used in test series II. The sharpened pins on the clamp faces engage the vertebral body endplates

Fig. 4. Complete test set-up: materials testing system, mechanical clamp, 4 degree of freedom table with specimen and rotationally unconstrained attachment jig

mentation [5, 6] and the Universal Spine System (Synthes). Several laminar and pedicular hooks or pedicle screws are placed at multiple levels along the spinal deformity. In addition, distribution of the corrective forces is meant to avoid stress concentration on only a few laminae or pedicles and therefore reduce the failure rate of the instrumentation.

However, hook disengagement has been reported with the CD system [9, 24, 26]. Clinical experiences with these implants in thoracic scoliosis surgery indicate that considerable forces may be applied to the hooks intra-operatively during three-dimensional correction of the vertebrae. In order to withstand these forces, pedicle hooks have been developed using various types of screws for additional fixation.

Although the pull-out strength of various pedicle screws has been investigated in many reports [10, 15, 16, 18, 23, 25, 28–30], few data are available on how much force can be applied to pedicle and lamina hooks and how they compare to pedicle screws. The aim of this study was to investigate the loading tolerances of three hooks which are components of posterior fixation systems for the thoracic spine: the USS pedicle hook, a prototype pedicle hook and the CD pedicle hook. A standard 5-mm USS pedicle screw was tested for comparison. In order to perform a clinically relevant test, the major loading force acting intraoperatively on the implants is simulated with constrained and rotationally unconstrained posterior pull-out tests.

Materials and methods

Implants

The hooks tested in this study are designed to provide rigid fixation to the posterior elements. The CD hook is secured using a 5.5-mm-long screw of 3-mm diameter (Fig. 1). This screw is tightened against the posterior facet surface. No additional holes need to be prepared. In contrast, the USS hooks use an additional, 3.2-mm, self-tapping fixation screw (length 25-mm or 30-mm) which passes obliquely, in the anterocranial direction, through the inferior facet of the vertebra into the pedicle (Figs. 1 and 2). The canal for the screw is prepared with a 2-mm drill. The prototype hook is additionally anchored to the ipsilateral pedicle of the caudally adjacent vertebra with a second 3.2-mm, self-tapping fixation screw requiring a second drilling procedure (Figs. 1 and 2). The self-tapping pedicle screw has a 5-mm major diameter, 3.8-mm minor diameter and 2-mm pitch (Fig. 1).

Spine preparation

Nine human thoracic spines (T2–12) with an average age of 55.8 years (ranging from 40 to 68 years) were used for this study. They were from one female and eight male cadavers. All spines were harvested within 24 h of death, frozen at –20°C and thawed for testing. Each spine was cleared of all soft tissues and disarticulated. The pedicle's sagittal and horizontal diameters were measured with a precision caliper. For pedicle screw implantation a 2-mm-diameter pilot hole was prepared with a hand drill. This hole was enlarged to 3.2 mm with a pneumatic drill, taking care not to damage the pedicular walls. The longest possible screw was inserted without perforating the anterior cortex, resulting in a 80%–100% insertion depth according to Krag [15].

The USS hook and the prototype hook were implanted into the facet joint after removal of the caudal portions of the inferior facet of the vertebra. A fixation screw was inserted into a 2-mm hole drilled obliquely through the remaining facet into the pedicle. The tip of the selected screw was placed as close to the superior pedicular cortex as possible. The prototype hook was additionally anchored to the facet and the ipsilateral pedicle of the caudally adjacent vertebra with a second fixation screw.

The CD hook was implanted similarly, and its set screw was tightened against the superior portion of the facet joint.

The specimens were than X-rayed to check for correct positioning of the implants. Each specimen was fixed to a test table in a posterior-up position such that the posterior elements, the pedicles and the implant were exposed.

The vertebral bodies of the series I specimens were moulded in ostalloy after reinforcement screws were placed through the vertebral bodies (Billiton Witmetaal, Naarden, Netherlands). In series II, the specimens were fixed in a custom mechanical "clamp" (Fig. 3). The faces of the clamp applied cranial/caudal compression to the vertebral body endplates. Further anchorage of the vertebral body to the clamp was provided by 1-cm-long and 3-mm-diameter sharpened pins which penetrated the endplate of the vertebral body.

Testing protocol

Posterior pull-out tests were performed using Instron testing systems (test series I, model 1272 servo-hydraulic; test series II, model 4302 screw-driven; Instron Corp, High Wycombe, Bucks, UK). The testing mode was displacement controlled at a constant rate of 0.5 mm/s.

The pedicle screw was loaded along its axis. The hooks were all loaded perpendicular to the intended rod direction. The required specimen alignment was obtained using a 4 degree of freedom test table (Fig. 4). A custom attachment jig connected the implant to the load cell. In series I the jig was rigidly affixed to the load cell. In series II the rotational constraints were removed by installing a universal joint and an axial thrust bearing between the attachment jig and the load cell.

Force and displacement data were sampled 50 times/s. The failure load was defined as the maximum load recorded.

Bone densitometry was performed on isolated vertebrae of nine spines using the dual energy X-ray absorptiometry technique (DXA; Hologic QDR 1000, Hologic Inc., Waltham, Mass.). After removal of the soft tissues, the vertebrae were positioned anteroposteriorly on an acrylic plate. The latter compensates for soft tissue and improves spectral distribution of the X-rays. Scans were analysed according to the procedure recommended by the manufacturer (Spine analysis software version 4.47). Results were expressed as bone mineral density (BMD, g/cm^2).

Table 1. Matched paired tests performed in series I (constrained) and series II (rotationally unconstrained)

Series	Configuration	Site of testing	Number of matched pairs
IA	USS hook versus pedicle screw	Same vertebra	8
IB	USS hook versus prototype hook	Adjacent vertebra	8
IC	USS hook versus CD hook	Adjacent vertebra	8
IIA	USS hook versus CD hook	Adjacent vertebra	9
IIB	USS hook versus prototype hook	Adjacent vertebra	8

Data analysis

A matched pair experiment procedure was used: pedicle screws and USS hooks were compared on the pedicles of the same vertebra if a clean pull-out of the pedicle screw was observed. In any other case, implants were tested on adjacent levels because pedicular fracture was expected as the failure mode. For the purposes of the paired *t*-test analysis, both implants tested on adjacent levels and implants tested in a single vertebra were considered to be paired samples.

In the first part of this study, all fixture degrees of freedom were constrained. In the second part of the study, all rotational fixture degrees of freedom were unconstrained (Table 1).

Linear regression analysis was performed to ascertain the following correlations:

- Pedicle screw diameter with failure load
- USS hook failure load with cross-sectional area of the pedicle
- USS hook failure load with pedicle height (sagittal pedicle diameter)
- USS hook failure load with bone mineral density
- USS hook failure load with length of the additional fixation screw

Results

Series I

The results of all pull-out tests in this series are summarized in Table 2. The failure displacement is defined as the displacement at the failure load. Results of the matched paired *t*-tests are summarized in Fig. 5.

Series IA. For the pedicle screw a total of 10 tests were performed with an average pull-out force of 1646.1 N (SD 622.8 N). In 7 cases a clean pull-out of the screw was observed, in 3 cases the pedicle fractured. A typical force-displacement relationship for the pedicle screw is presented in Fig. 6. The sudden drop in force which occurs at approximately 2.5 mm displacement indicates screw pull-out.

Table 2. Summary of data obtained from all constrained tests (series I)

Implant	Pedicle screw	Prototype hook	USS hook	CD hook
Number of tests	10	9	25	8
Average failure load (N)	1646.1	1529.9	906.2	742.8
Maximal loads (N)	2729.6	2216.9	1406.3	930.2
Minimal loads (N)	961.0	954.6	522.5	449.2
SD of failure load (N)	622.8	414.0	243.0	189.5
Average failure displacement (mm)	2.4	2.2	2.1	2.9
SD of failure displacement (mm)	1.0	0.9	1.0	0.9

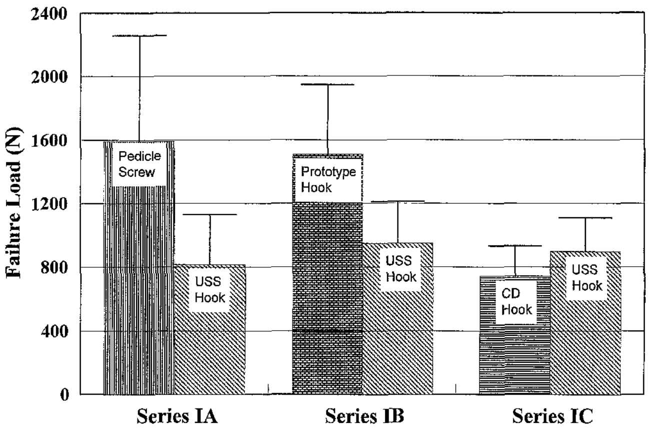


Fig. 5. Results of matched paired *t*-tests for constrained series I. The differences indicated for series IA and series IB are significant at the $P < 0.05$ level

The USS hook was tested nine times. An average pull-out force of 849.0 N (SD 309.4 N) was observed. The typical failure mode for the USS hook was a pedicular fracture along the fixation screw without any major movement of the hook relative to the bony implantation site. A typical force-displacement relationship for the USS hook is illustrated in Fig. 6. The sharp drops in force occurring at approximately 2.5 mm and 8.0 mm displacements are indicative of a pedicle fracture.

Results of the matched paired *t*-test indicated that the pedicle screw pull-out load was significantly greater than that for the pedicle hook at the $P < 0.05$ value.

Series IB. The prototype hook was tested nine times. An average pull-out force of 1529.9 N (SD 414.0 N) was measured. The failure mode was comparable to the standard USS hook. The additional fixation screw usually cut out cleanly. In three cases the caudal pedicle fractured. A typical force-displacement curve for the prototype hook is given in Fig. 6.

In this series a total of eight USS hooks were tested. An average pull-out force of 949.7 N (SD 261.0 N) was measured. Results of the *t*-test indicated that the prototype hook pull-out force was significantly greater than that for the USS pedicle hook at the $P < 0.05$ value.

Series IC. In eight tests the average pull-out force of the CD hook was 742.8 N (SD 189.5 N). The failure mode observed was an initial cranial translation and deformation of the posterior elements followed by a fracture of the facet and lamina. A typical force-displacement curve for the CD hook is presented in Fig. 6. The sharp decrease in force occurring at approximately 1.6 mm of hook displacement is indicative of a facet/lamina fracture.

Eight USS hooks were tested in this series. An average pull-out force of 896.3 N (SD 213.1 N) was measured. Results of the *t*-test indicated the difference between the CD and USS hooks at the $P < 0.05$ value was not significant.

Series II

The results of all pull-out tests in this series are given in Table 3. The nine sample paired *t*-test is summarized in Fig. 8.

Series IIA. The USS hook and the CD hook were both tested nine times. The USS hook failed at an average pull-out force of 803.6 N (SD 147.2 N). In all cases the failure mode was a fracture of the pedicle, occurring seven times along the axis of the fixation screw and twice at the pedicle/vertebral body junction. The USS hook did not translate from its bony anchorage in any test. A typical force-displacement curve for the USS hook is given in Fig. 7.

For the CD hook an average pull-out force of 463.7 N (SD 279.1 N) was observed. The results were considerably more scattered than for the USS hook. The typical failure mode was a cranial translation along the facet and a slip off from the lamina without fracturing the bone. A typical force-displacement curve for the CD hook is presented in Fig. 7. The matched paired *t*-test indicated that the USS hook pull-out load was significantly greater than that for the CD hook at the $P < 0.05$ value.

Series IIB. The USS hook and the prototype hook were both tested eight times. For the USS hook an average pull-out force of 778.2 N (SD 228.1 N) was observed. Again the typical failure was a pedicular fracture along the axis of the fixation screw without movement of the hook itself.

The prototype hook failed at an average load of 1617.1N (SD 652.3 N); the maximum failure load was 2690.6 N. In six cases a clean pull-out of the second screw was observed, in two cases the caudal pedicle fractured. A typical force-displacement curve for the prototype hook is presented in Fig. 7. The matched paired *t*-test indicated that the prototype hook pull-out load was significantly greater than that for the USS hook at the $P < 0.05$ value (Fig. 8).

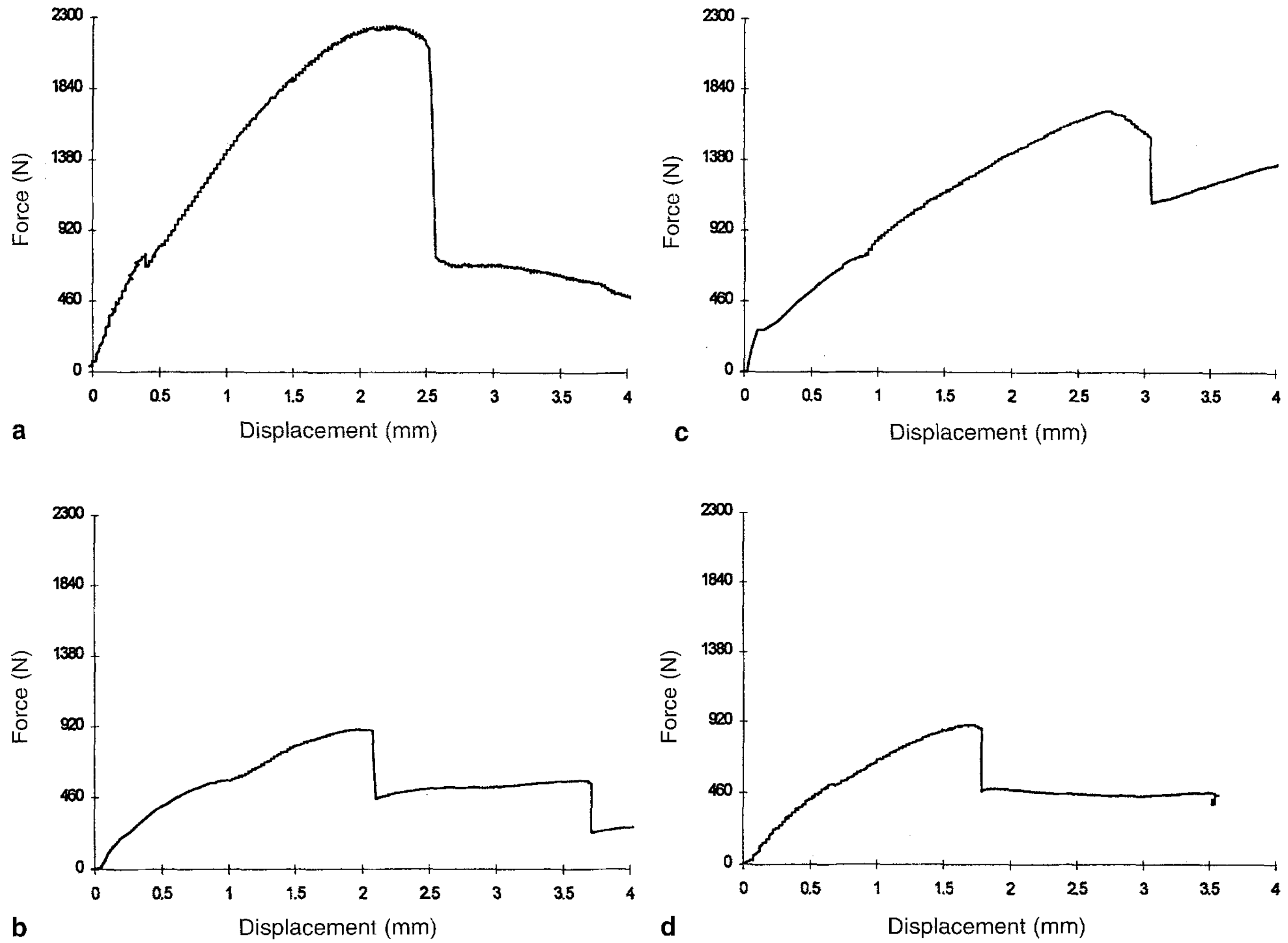


Fig. 6. Typical force-displacement curves for the constrained series I: **a** pedicle screw, **b** USS pedicle hook, **c** prototype pedicle hook and **d** CD pedicle hook

Pedicle dimensions

Transverse and sagittal pedicle diameters are given in Table 4. The lowest average pedicle width was observed at the T5 level. The lowest average pedicle height was measured at the T6 level.

Bone mineral density

A total of nine reference vertebrae were measured. BMD values are given in Table 5.

Regression analyses

There was no statistically significant correlation of the pedicle dimensions and the failure loads of any implant tested. A pedicle screw failure load of more than 1800 N was recorded in three of four tested upper thoracic vertebrae. However, this trend was of no statistical significance.

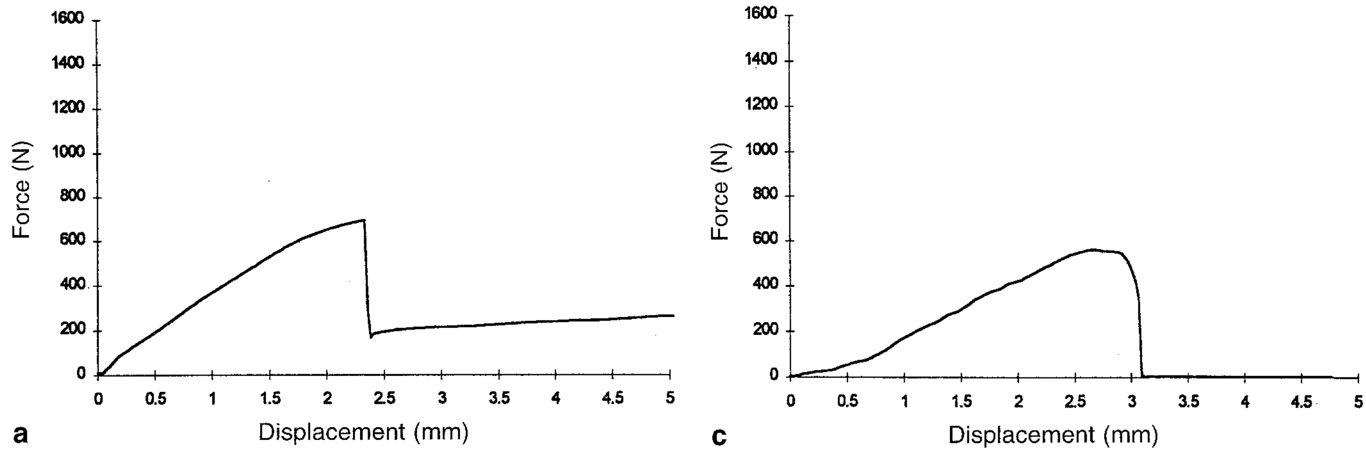
Further analysis showed that the failure load of the USS hooks was not correlated to the length of the additional fixation screw. The BMD did not correlate with the failure load of the USS hook.

Table 3. Summary of data obtained from all unconstrained tests (series II)

Implant	Prototype hook	USS hook	CD hook
Number of tests	8	17	9
Average failure load (N)	1617.1	791.6	463.7
Maximum observed load (N)	2690.6	1174.6	939.2
Minimum observed load (N)	548.1	489.5	91.4
SD of failure load (N)	652.3	183.8	279.1
Average failure displacement (mm)	3.1	2.6	1.6
SD of failure displacement (mm)	1.0	1.1	1.0

Discussion

Pedicle and lamina hooks are widely used in posterior thoracic spine surgery. Rotations of the hook with respect to the lamina, dislodgement of the hook from the lamina and even fracture of the posterior elements have been reported in clinical studies [15, 22]. Double hooks and locking hooks have been developed in order to reduce the



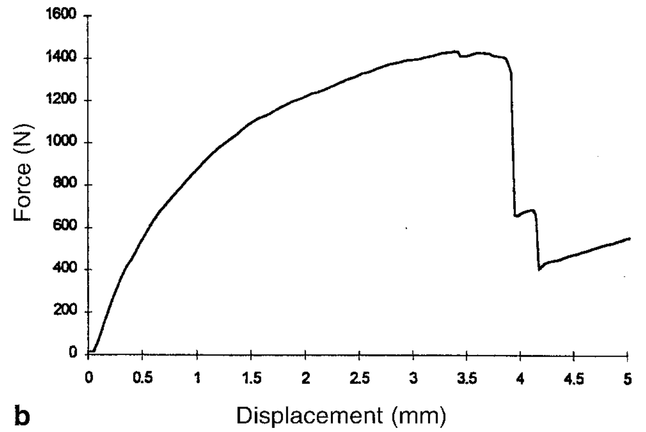


Fig. 7. Typical force-displacement curves for the unconstrained series II: **a** USS pedicle hook, **b** prototype pedicle hook and **c** CD pedicle hook (pedicle screw not evaluated in series II)

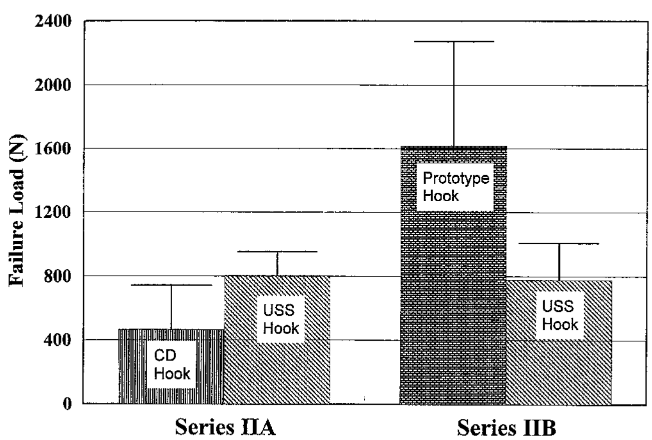


Fig. 8. Results of matched paired *t*-tests for unconstrained series II. The differences indicated for both series IIA and IIB are significant at the $P < 0.05$ level

Table 4. Pedicle dimensions in nine thoracic spines (mean and SD)

Level	Transverse diameter (mm)		Sagittal diameter (mm)	
	Mean	SD	Mean	SD
T12	8.9	1.7	17.4	1.8
T11	8.7	1.2	17.6	1.9
T10	7.9	1.2	16.5	1.4
T9	7.3	1.2	15.0	1.2
T8	6.6	1.2	13.6	1.8
T7	6.1	1.2	12.8	1.0
T6	5.8	0.9	12.1	1.1
T5	5.4	0.7	12.1	1.6
T4	5.9	0.8	12.3	1.4
T3	6.2	1.3	12.5	1.5
T2	6.5	0.6	12.0	1.4

number of these failures [2, 13]. However, transmission of three-dimensional corrective forces onto the vertebra requires additional fixation. All three hooks tested are additionally instrumented with fixation screws (Fig. 1).

Several studies have been performed to investigate the pull-out forces of single screws [25, 29] as well as the biomechanical properties of complete fixation systems [7, 19, 20]. Relatively few investigators have compared these findings with pedicle hook pull-outs [3, 21]. Therefore, we tested the pull-out force of the 5-mm USS pedicle screw in addition to the pedicle hooks.

Biomechanical testing

In order to evaluate the pedicle hooks and pedicle screws biomechanically, posterior pull-out tests have been pro-

Table 5. Results of the bone mineral density measurements. The average values were regarded as representative for the complete spine. Spines no. 4–10 were used for the regression analysis

Spine no.	Verebra(e) measured	(Average) BMD (g/cm^2)
1	T3–4	0.714
3	T8	0.643
4	T12	0.854
5	T10–11	0.988
6	T12	0.835
7	T12	0.767
8	L1	0.880
9	T11–L1	0.879
10	T12	0.844

posed [3, 21]. To simulate intraoperative conditions, constrained (series I) and rotationally unconstrained (series II) pull-out tests were performed. The applied force in series I corresponds to the situation found with the CD technique wherein the rod is rotated by 90° to correct flexible scoliotic deformities. The loading of the hooks is considered "constrained", since the implants are always aligned with the rod. In the USS case, the vertebrae may be individually connected to the rod, allowing so-called "segmental correction". The instrument used intraoperatively to perform this correction leaves the rotational degrees of freedom unconstrained. This situation was simulated in test series II by inserting a universal joint between the implant and the load cell.

The pedicle screw was the strongest device, with a maximum pull-out force of 2729.6 N (Table 2). In three cases involving small pedicles, the failure mode observed was a fracture of the pedicle rather than a clean pull-out of the thread. The importance of the "cortical cylinder" surrounding the pedicle for the anchorage of pedicle screws has been stressed in other studies [18, 30]. The thickness of the cortical cylinder has been reported to range from 15% to 40% of the pedicle's minor dimension [18]. This suggests a higher pedicle screw failure load for smaller pedicles as the screw threads may engage the cortical bone. Moran et al. [18] could not confirm this, and we observed a statistically insignificant trend only.

For stabilization of the USS hook, the additional screw fixation to the pedicular cortical bone at the screw's entry point seemed to be the major factor. Of the 42 USS hooks tested, only three slipped slightly caudally. This stable fixation led to a constant failure pattern: a pedicle fracture along the axis of the inserted screw. The average failure load of the USS hook was 55% of that for the pedicle screw in the constrained test. This failure load did not change significantly under rotationally unconstrained conditions. The length of the fixation screw (25 or 30 mm) and therefore the amount of engagement of cancellous bone within the pedicle did not influence the hook's loading tolerance.

The fixation strength of the USS hook might be further improved if the fixation screw perforated the strong bone of the vertebral endplate (J. K. Webb, 1994, pers. comm.). However, in the hook configuration tested a longer fixation screw would have missed the endplate posteriorly (Fig. 2). The angulation of the fixation screw has subsequently been modified so that its trajectory now perforates the endplate. The biomechanical evaluation of these modified hooks is currently under way at our institute.

The average failure load of the CD hook was 45% of that for the pedicle screw in the constrained test. This failure load did decrease significantly under rotationally unconstrained conditions to 28% of that for the pedicle screw. In addition, the range of failure loads exhibited considerable scattering (91.4 – 939.2 N) and may have resulted from the variation in the fit between the hook and the facet joint. In contrast to the USS hook, the CD hook failed by translating away from its original position. This movement was limited in the constrained test, leading to a fracture of the lamina. With additional degrees of freedom the hook failed, in all cases, by suddenly slipping completely off the facet joint, suggesting a mechanical instability.

Ruland et al. [21] reported average failure loads of 809 N (SD 99 N) for the CD laminar hook, and Coe et al. [3] found a posterior cutting-out through the lamina as the most common failure mode for these devices. However, details of the degree of constraint imposed on the implants was not reported for these studies. Freedman et al. [8] investigated Harrington hooks by applying increasing distraction force and found loading tolerances between 660 and 1100 N. A fracture of the lamina in all trials may have been caused by the highly constrained nature of this test.

The prototype hook was developed to improve the fixation strength of the standard USS hook by using a second fixation screw which passes through the caudal "heel" of the hook (Fig. 1) into the facet and the pedicle of the caudally adjacent vertebra. The prototype hook was associated with significantly higher failure loads than the standard USS hook. The additional fixation screw at the heel increased the average fixation strength of the USS hook from 55% to 93% of the loading tolerance of the pedicle screw under constrained and from 48% to 98% under unconstrained conditions. All hooks failed by fracturing the pedicle without moving caudally from their site of implantation. However, this device bridges a spinal motion segment by connecting two adjacent vertebrae, hindering segmental spinal manipulation and correction. In vivo, the additional fixation screw may also be subjected to high loads during motion of the spinal segment, possibly resulting in its failure. Furthermore, the implantation of two additional fixation screws into the posterior elements and the pedicles exposes the patient to the risk associated with this procedure.

Implantation

All devices are in some respect limited in their use by anatomical features. The pedicle screw, the strongest implant, could not be inserted safely into the very small pedicles of the upper and middle thoracic region. Transverse pedicle diameters as low as 3.5 mm were measured (Table 4), which contradict the use of a pedicle screw but not of a pedicle hook. In the lower thoracic regions the hooks are limited by the shape, orientation and thickness of the facet joint. In addition, in the case of large diameters of the pedicles, the hook's bifid blade is sometimes too small to fit the pedicle properly. In our study the lowest level that hooks could be implanted was T10.

Bone mineral density

An et al. [1] reported a linear correlation of the BMD with anterior vertebral screw pull-out strength. Coe et al. [3] found that in posterior pull-out tests, failure loads for laminar hooks were independent of the BMD. In contrast, they showed a correlation of BMD and pedicle screw pull-out forces and therefore concluded that, in osteoporotic patients, hooks might be more effective than pedicle screws. We can confirm their findings regarding the independence of hook failure load from BMD, but our data do not allow conclusions concerning the relation between pedicle screw pull-out forces and BMD.

Test set-up

High loads initially led to complications with the fixation of the vertebra to the testing table: the vertebral body fractured along the axis of the reinforcement screws, and subsequently pulled out of the moulding material on three occasions. These tests were discarded since the failure was not related to the implant.

The method of affixing the vertebral body to the testing machine with a mechanical clamp was developed in response to the vertebral body fractures. The mechanical clamp was thought to distribute the pull-out load over the complete vertebral body endplate instead of concentrating the load at the location of the reinforcing screws. Although two vertebral body fractures occurred when the mechanical clamp was used, this device did provide a significant saving in time and costs compared with the moulding techniques.

Conclusions

The results of this study suggest that additional screw fixation improves the attachment of pedicle hooks to their implantation site. Hooks using fixation screws passing obliquely through the facet into the pedicle apparently provide the best attachment. Due to the firm anchorage to the posterior elements, the hooks with an oblique fixation screw may be subjected to three-dimensional loads without the danger of immediate dislodgement. This should allow the intraoperative application of manipulative correcting forces on each individual vertebra, enabling segmental spinal correction. The second fixation screw of the prototype hook almost doubles the pull-out strength of the USS hook. Although the fixation screws have to be drilled through the pedicle, the prototype hook might be considered an alternative to the pedicle screw, especially with small pedicle dimensions as found in the middle and upper thoracic region.

References

1. An HS, Lim TH, Enanich C, Hasanoglu KY, McGrady L (1993) Strength of the anterior vertebral screw fixation in relation to bone mineral density. BED Vol 24, Bioengineering Conference ASME, pp 469–471
2. Benzel EC (1993) Short-segment compression instrumentation for selected thoracic and lumbar spine fractures: the short-rod/two claw technique. J Neurosurg 79:335–340
3. Coe JD, Warden KE, Herzig MA, McAfee PC (1990) Influence of bone mineral density on the fixation of thoracolumbar implants – a comparative study of transpedicular screws, laminar hooks and spinous process wires. Spine 15:902–907
4. Cook SD, Barrack RL, Georgette FS, Whitecloud TS, Burke SW, Skinner HB, Renz EA (1985) An analysis of failed Harrington rods. Spine 10:313–316
5. Cotrel Y, Dubousset J (1984) A new technique of spine fixation by a posterior approach in the treatment of scoliosis (in French). Rev Chir Orthop 70:489–494
6. Cotrel Y, Dubousset J, Guillaumat M (1988) New universal instrumentation in spinal surgery. Clin Orthop 227:10–23
7. Eggli S, Schläpfer F, Angst M, Witschger P, Aebi M (1992) Biomechanical testing of three newly developed transpedicular multisegmental fixation systems. Eur Spine J 1:109–116
8. Freedman LS, Houghton GR, Evans M (1986) Cadaveric study comparing the stability of upper distraction hooks used in Harrington instrumentation. Spine 11:579–582
9. Guidera KJ, Hooten J, Weatherly W, Highhouse M, Castellvi A, Ogden JA, Pugh L, Cook S (1993) Cotrel-Dubousset instrumentation – results in 52 patients. Spine 18:427–431
10. Halvorson TL, Kelly LA, Whitecloud TS (1991) Transpedicular screw pullout strengths in normal and osteoporotic human spines. Trans Orthop Res Soc 16:24
11. Harrington PR (1962) Treatment of scoliosis. J Bone Joint Surg [Am] 44:591–610
12. Harrington PR, Dickson JH (1973) An eleven-year clinical investigation of Harrington instrumentation – a preliminary report on 578 cases. Clin Orthop 93:113–130
13. Jacobs RR, Schläpfer F, Mathys R Jr, Nachemson A, Perren SM (1984) A locking hook spinal rod system for stabilization of fracture-dislocations and correction of deformities of the dorsolumbar spine: a biomechanic evaluation. Clin Orthop 189:168–177
14. Karjalainen M, Aho AJ, Katevuo K (1992) Operative treatment of unstable thoracolumbar fractures by the posterior approach with the use of Williams plates or Harrington rods. Int Orthop 16:219–222
15. Krag MH (1991) Biomechanics of thoracolumbar spinal fixation. Spine 16(3S):84–99
16. Krag MH, Beynnon BD, Pope MH, DeCoster TA (1989) Depth of insertion of transpedicular vertebral screws into human vertebrae: effect upon screw-vertebra interface strength. J Spinal Disord 1:287–294
17. Lagrone MO, Bradford DS, Moe JH, Lonstein JE, Winter RB, Ogilvie JW (1988) Treatment of symptomatic flatback after spinal fusion. J Bone Joint Surg [Am] 70:569–580
18. Moran JM, Berg WS, Berry JL, Geiger JM, Steffee AD (1989) Transpedicular screw fixation. J Orthop Res 7:107–114
19. Nolte LP, Steffen R, Krämer J, Jergas M (1993) The fixateur interne: a comparative biomechanical investigation (in German). Akt Traumatol 23:20–26
20. Panjabi MM, Yamamoto I, Oxland TR, Crisco JJ, Freedman D (1991) Biomechanical stability of five pedicle screw fixation systems in a human lumbar spine instability model. Clin Biomech 6:197–205
21. Ruland CM, McAfee PC, Warden KE, Cunningham BW (1991) Triangulation of pedicular instrumentation – a biomechanical analysis. Spine 16(6S):270–276
22. Sasso RC, Cotler HB (1993) Posterior instrumentation and fusion for unstable fractures and fracture-dislocations of the thoracic and lumbar spine – a comparative study of three fixation devices in 70 patients. Spine 18:450–460

23. Sell P, Collins M, Dove J (1988) Pedicle screws: axial pullout strength in the lumbar spine. Spine 13:1075–1076
24. Shufflebarger HL, Clark CE (1990) Complications of CD in idiopathic scoliosis. Presented at the combined meeting of the Pediatric Orthopaedic Society of North America and the European Pediatric Orthopaedic Society, Montreal, Canada
25. Skinner R, Maybee J, Transfeldt E, Venter R, Chalmers W (1990) Experimental pullout testing and comparison of variables in transpedicular screw fixation – a biomechanical study. Spine 15:195–201
26. Van Ooy A, Geukers CWGM (1992) Results of CD operation in idiopathic scoliosis. Acta Orthop Belg 58(S1):129–133
27. Willers U, Hedlund R, Aaro S, Normelli H, Westman L (1993) Long-term results of Harrington instrumentation in idiopathic scoliosis. Spine 18:713–717
28. Wittenberg RH, Shea M, Swartz DE, Lee KS, White AA III, Hayes WC (1991) Importance of bone mineral density in instrumented spine fusions. Spine 16:647–652
29. Wittenberg RH, Lee KS, Shea M, White AA III, Hayes WC (1993) Effect of screw diameter, insertion technique, and bone cement augmentation of pedicular screw fixation strength. Clin Orthop 296:278–287
30. Zindrick MR, Wiltse LL, Widell EH, Thomas JC, Holland WR, Field BT, Spencer CW (1986) A biomechanical study of intrapeduncular screw fixation in the lumbosacral spine. Clin Orthop 203:99–112