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ORIGINAL ARTICLE

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Primary stability of anterior lumbar stabilization: interdependence of implant type and endplate retention or removal

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Abstract This is a comparative in vitro biomechanical study of the primary stability of an anterior lumbar interbody stabilization. The objective was to compare the stability of a interbody stabilizing titanium cage with and without the retention of the bordering vertebral endplates, as well as to compare the titanium cage with a tricalcium phosphate block when the endplates are removed. An adequate blood supply is critical for interbody fusion, which suggests surgical treatment of the bordering endplates. On the other hand, primary stability is improved by the retention of the endplates. Furthermore, bone substitute materials are finding more frequent use due to complications associated with autologous bone grafts. Ten bovine lumbar spine motion segments (average age 6 months) were investigated. Pure bending loadings as well as eccentric axial compression loadings were applied. A titanium cage and tricalcium phosphate block, were tested in conjunction with an anterior augmentation (MACS). Range of motion, neutral zone (NZ) and bending stiffness were measured under pure bending to 10 Nm, and bending stiffness under axial loads of up to 1500 N. Range of motion of both implants in flexion-extension was significantly smaller than physiologic (cage without endplates 4.3°, cage with 2.8°, block without 3.4°, and physiologic 6.6°, all $p < 0.001$). The cage with endplates and the block without endplates were both significantly stiffer than physiologic in all directions except left lateral bending. The block without endplates and the cage with endplates were both stiffer than the cage without endplates. The results suggest that the use of the bone substitute block provides better stability than the cage when the endplates are removed.

Keywords Biomechanical properties · Endplates · Anterior instrumentation · Ceramic block

Introduction

Fusion surgeries of all regions of the vertebral column are increasing steadily in number. In the year 2001 alone, approximately 360,000 arthrodeses were performed on the spine in the USA [2]. The aim of spinal fusion is to achieve bony stiffening of the affected segment. Generally, this goal has been achieved in practice by performing a dorsal fusion (PLF), a dorsal interbody fusion (PLIF), a ventral interbody fusion (ALIF) or a combined fusion [7, 29, 67]. Decisive for success of the fusion is an uninterrupted consolidation zone, which suggests the importance of surgical treatment of the bordering upper and lower endplates. The greatest structural stability of the vertebral body is located at the endplates [21, 62]. The sacral endplates are more stable than the lumbar endplates, the lower more stable than the upper endplates, and the posterior sections more stable than the ventral sections of

the endplates [18]. The more cortical bone that is preserved in the lower and upper endplates, the more primary stability is ensured during osteosynthesis, and the lower the likelihood that interbody implants such as cages or blocks are going to shift [13, 18, 21, 31]. Nevertheless, the joining of the implant to the blood supply of the vertebral body is essential for bony fusion to occur. The more of the upper and lower endplates that is removed, the greater the chances of obtaining an optimal union.

In the 1960s, common practice was to remove the endplates completely which led to inadequate stability and the shifting of the implants [29]. Later, the endplates were fully preserved and only the cartilaginous surface removed, however, bony fusion was not always achieved [29]. Comparable biomechanical studies regarding the preservation or conversely the complete removal of the upper lower endplates in ALIF have not been published to date.

Due to the high rate of complications of up to 25% observed during removal of autologous bone from the iliac crest, bone substitute materials are being used more frequently in fusion operations [3, 49]. The use of bone substitute eliminates many of the major and minor complications arising on removal of bone, however, the biological and mechanical problems of associated with bone substitute materials remain unsolved. In the literature, several investigations have been described concerning osteointegration of bone substitute materials in the vertebral column in a number of animal models [9, 14, 16, 39, 60]. Furthermore, clinical reports on the biological reaction of ceramic material in the human vertebral column have been published [32, 54, 59]. On the other hand, very little has been mentioned with regards to the biomechanical properties of these bone substitute materials in providing mechanical stability during spinal fusion.

It has been observed that ceramic materials may sometimes fracture, either intraoperatively or within the first few months postoperatively. However, this apparently results in no change in the position of the treated segment [9, 60]. This study was conducted to address the questions: what is the role of the bordering upper and lower endplates in the primary stability of ventral intercorporal stabilization, and can the primary stability of a ventrally instrumented segment be improved by replacing the normally used cage with a suitable ceramic material.

Materials and methods

Ten (10) motion segments from six bovine lumbar spines were used for these in vitro investigations. The animals were 6-month old at the time of slaughter and weighed on average 146 kg (130–159 kg). Motion segments L1/L2 and L3/L4, respectively, were taken from the lumbar spines for testing. Care was taken to completely preserve the bony and ligamentous structures of the locomotor segment on preparation of the specimens, with removal only of muscular and fatty tissue. Following preparation, the specimens were stored frozen at −20°C, and thawed at room temperature for 12 h prior to testing. The cranial and caudal ends of the vertebral bodies of the motion segments were embedded in cold curing methylmethacrylate resin (Technovit 4004, Heraeus Kulzer, GmbH, Wehrheim, Germany) in special molds designed for this purpose.

Two different systems were used to biomechanically evaluate the specimens, a multi-axis pure moment apparatus (PMA), and an uniaxial material test machine, (Minibionix 858, MTS Corporation, Minneapolis, MN, USA) which was used to apply combined axial-compression and bending moment loading. All specimens were examined using both testing systems. The PMA for non-destructive stability testing of spine motion segments was constructed according to the principles established by Panjabi [41]. It applies pure moment loadings to the specimens without imposing axial or shear force loading, and can apply loadings in the sagital, frontal, as well as axial planes of the motion segment (Fig. 1). Moment loading was applied without preload by means of a cable and disc arrangement in ten increments to10 Nm in flexion/extension and lateral bending,

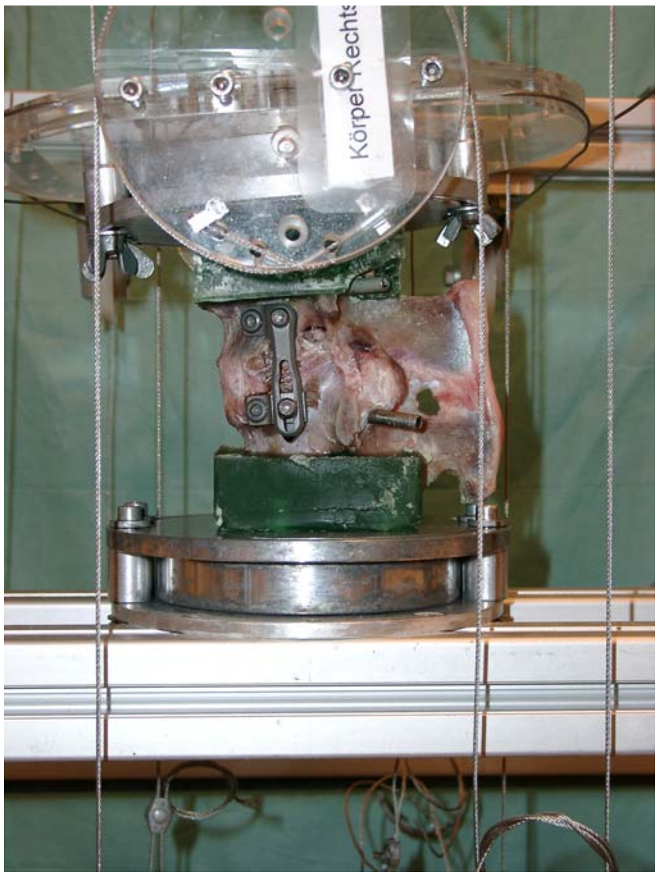


Fig. 1 Pure moment apparatus (*PMA*)

and in six increments to 4 Nm in axial rotation. Motion of the specimen was measured using an ultrasound based motion analysis system with a resolution of 0.1° (CMS100, Zebris Medizintechnik GmbH, Isny, Germany). The MTS testing applies combined axial and moment loading to the motion segment. These combined loadings were achieved by applying an axial force at a defined offset to the superior-inferior axis of the specimen, loading was applied at a rate of 25 N/s.

The implant system tested was the modular anterior construct system (MACS)-System which is designed as a ventral plate system for the treatment of thoracolumbar vertebral column (Aesculap A& Company, KG, Tuttlingen, Germany). This implant is manufactured from a titanium alloy and can be inserted both using an open as well as endoscopic surgical approach. The plate has a low profile (<10 mm). All implants were performed by the same surgeon in accordance with guidelines provided by the manufacturer. The Pyramesh cage was used as a stabilizing device to fill the intervertebral disk space to a height corresponding to the level of the intervertebral disks (almost 1 cm) (Medtronic Sofamor Danek, Memphis, TN, USA). The cages were trimmed to the appropriate size from cages having an original length 30 mm and diameter of 19 mm using the same technique that is used intraoperatively. In the specimens that were tested without upper and lower endplates, higher cages were necessary and these were also cut to a height of about 2 cm. Biosorb, a bone substitute material (Science for BioMaterials, Lourdes, France) was used as an alternative to the large cages after removal of the intervertebral disk and the complete upper and lower endplates using an osteotome, so that cancellous bone was exposed. The material consisted of a synthetically produced tricalcium phosphate with a porosity of around 30%, a compressive strength of 30 MPa, and a pore size of 250–400 µm. Blocks were trimmed to 30 mm in length so that they fit snugly into the prepared intervertebral space. The width of the blocks used was 20 mm, the height of the blocks was dependent on the specimens and corresponded approximately to that of the large cages (approximately 2 cm). Prior to implanting, the height of the mounting blocks into which the vertebral bodies were embedded was measured. This distance was maintained for all implants and additional compression of the motion segments was rendered unnecessary.

The motion segment was first tested in its physiological state, after which the MACS plate was attached and the tests repeated. The plate was then removed leaving the accompanying screws in position. The intervertebral disk was resected taking care not to impair the ligamentum longitudinal anterius from an antero-left-lateral approach corresponding to the intraoperative technique used from L1 to L5. The anulus fibrosus on the opposite side was preserved, and the small Pyramesh cage implanted, the MACS-plate reattached and the motion segment tested. The upper and lower endplates were then removed totally, taking care not to damage the anterior and posterior longitudinal ligament or opposite anulus fibrosis. A large Pyramesh cage was then inserted, the MACS-plate reattached, and the motion segment tested again. Finally, the Biosorb bone-block was implanted and also tested with the MACS-plate in place. It was shown in preliminary experiments that removal and reattachment of the MACS-plate did not affect the stability of the motion segment.

All specimens were X-rayed along two planes following preparation and each completed test in order to verify the correct position of the implant and to exclude any bony damage to the specimens. Range of motion, neutral zone (NZ), and bending stiffness (summarized in Table 1) were measured on the ten specimens with each plate-implant combination described above.

With the PMA, the left and right axial rotation, flexion/extension, and left and right lateral bending were tested to a maximum moment of 10 Nm. Motion of the specimens was measured using an ultrasound-based 3D motion analysis system (CMS–100, Zebris GmbH, Isny, Germany). Using the MTS, flexion was produced by applying an axially directed force at a 20 mm offset to the center of the intervertebral disk (Flex20). The axial force was applied up to a maximal of 1500 N. The same procedure was used to test extension at 20 mm offset (Ex20), and lateral bending at a 30 mm offset (LBl30, LBr30). Motion of the specimen was measured with an LVDT (MLT 25, Data Instruments, Acton, USA) calibrated to measure angular motion. The apparatus was programmed to switch off automatically in the case that angular motion exceeded 5° in order to avoid any permanent deformations and damage to the specimens. Radiological documentation of specimens was carried out before and after these investigations.Evaluation, statistical assessment, and graphical representation of the measured data were carried out with the programs Origin 4.0 (Microcal Software Inc., Northhampton, USA), Excel (Microsoft Corporation, Bellevue, USA) and SPSS 10.0 (SPSS Inc. Chicago, USA). Average values and standard deviations were determined for all

Table 1 Parameters, method, and direction tested

Parameter tested	Direction tested	Testing method
Range of motion (ROM)	Flexion-extension Lateral bending Axial rotation	PMA
Neutral zone (NZ)	Flexion-extension Lateral bending Axial rotation	PMA
Bending stiffness (BES)	Flexion-extension Lateral bending Axial rotation	PMA
Bending stiffness (BES)	Flexion-extension Lateral bending	MTS

measured and calculated values. Normal distribution of the data was verified using the Kolmogorow–Smirnow-Test. Means were compared using the single factor analysis of variance ANOVA. In the case of significance, differences between groups were tested using the Least Significant Difference (LSD)-Test [55]. The significance level for measured value differences was $\alpha < 0.05$.

Results

The three instrumentation methods (MACS with small cage; MACS with large cage, and MACS with block) were tested in this study in comparison with the MACS-plate alone and with the uninstrumented physiologic specimens. From the moment-angle diagrams of the four instrumentation methods as well as the physiologic specimens tested with the PMA, it can be seen that the physiologic specimens exhibit considerably greater range of motion (ROM) than the instrumented segments (Fig. 2a). These exhibited very similar curves with exception of the group with the larger cage, the movement of which was larger when compared to the other three-instrumented groups. Similar behavior was observed in lateral bending (see Fig. 2b) as in flexion/extension. Of note here is that the range of motion of

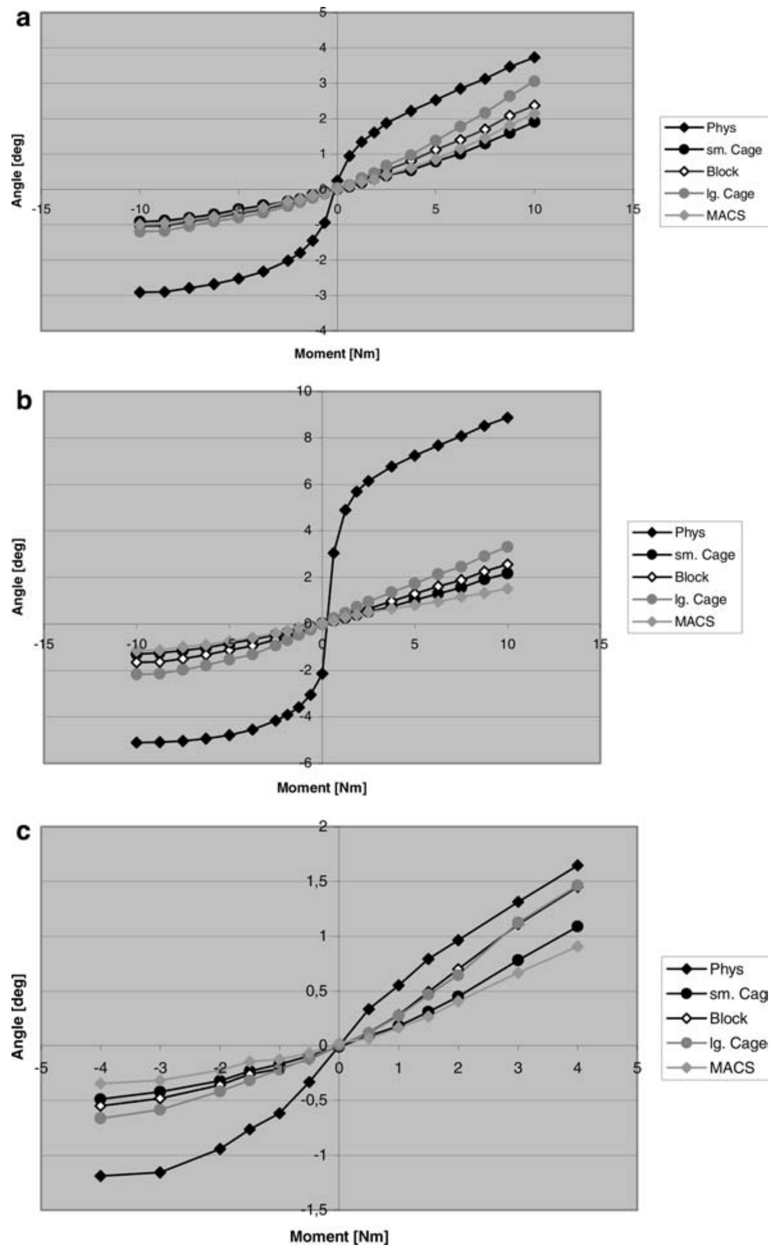


Fig. 2 a Moment-angle diagram (*Extension left*, *Flexion right*) from tests conducted with the PMA. Curves are average values of the $N = 10$ specimens tested per group. **b** Moment-angle diagram for lateral bending with the PMA (bending to the *left* is represented on the left-hand side, bending to the *right* on the right-hand side of the diagram). Curves are average values of the $N = 10$ specimens tested per group. **c** Moment-angle diagram for axial rotation with the PMA (rotation to the *left* on the left, rotation to the *right* on the right-hand side of the diagram). Curves are average values of the $N = 10$ specimens tested per group

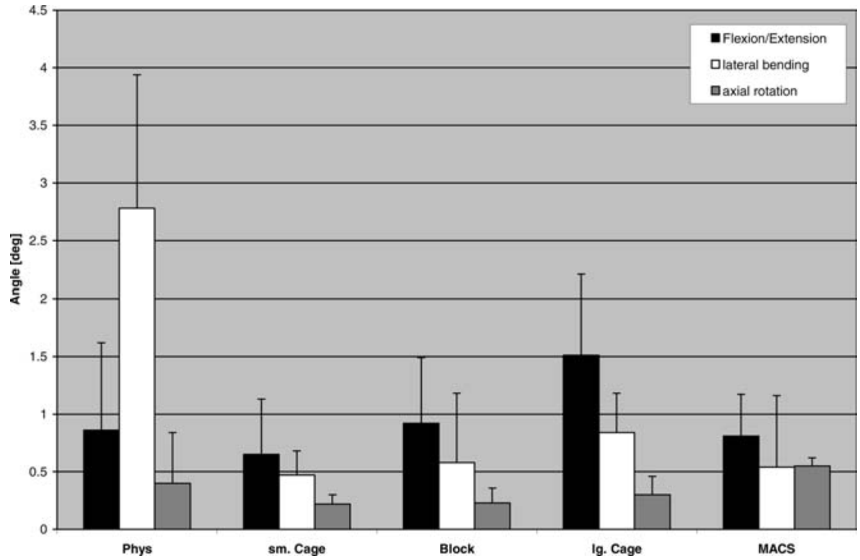


Fig. 3 Natural zone (*NZ*) for flexion/etension, lateral belonging abd axial rotation with the PMA

the physiologic specimens was markedly larger than that of the instrumented groups and, within these groups again, the range of motion was highest in the group with the larger cage (Fig. 1b). In axial rotation, all range of motion values were again similar in trend compared to the other two directions of motion although of a lower magnitude, and the range of motion of the physiologic specimens was greatest here as well (Fig. 2c).

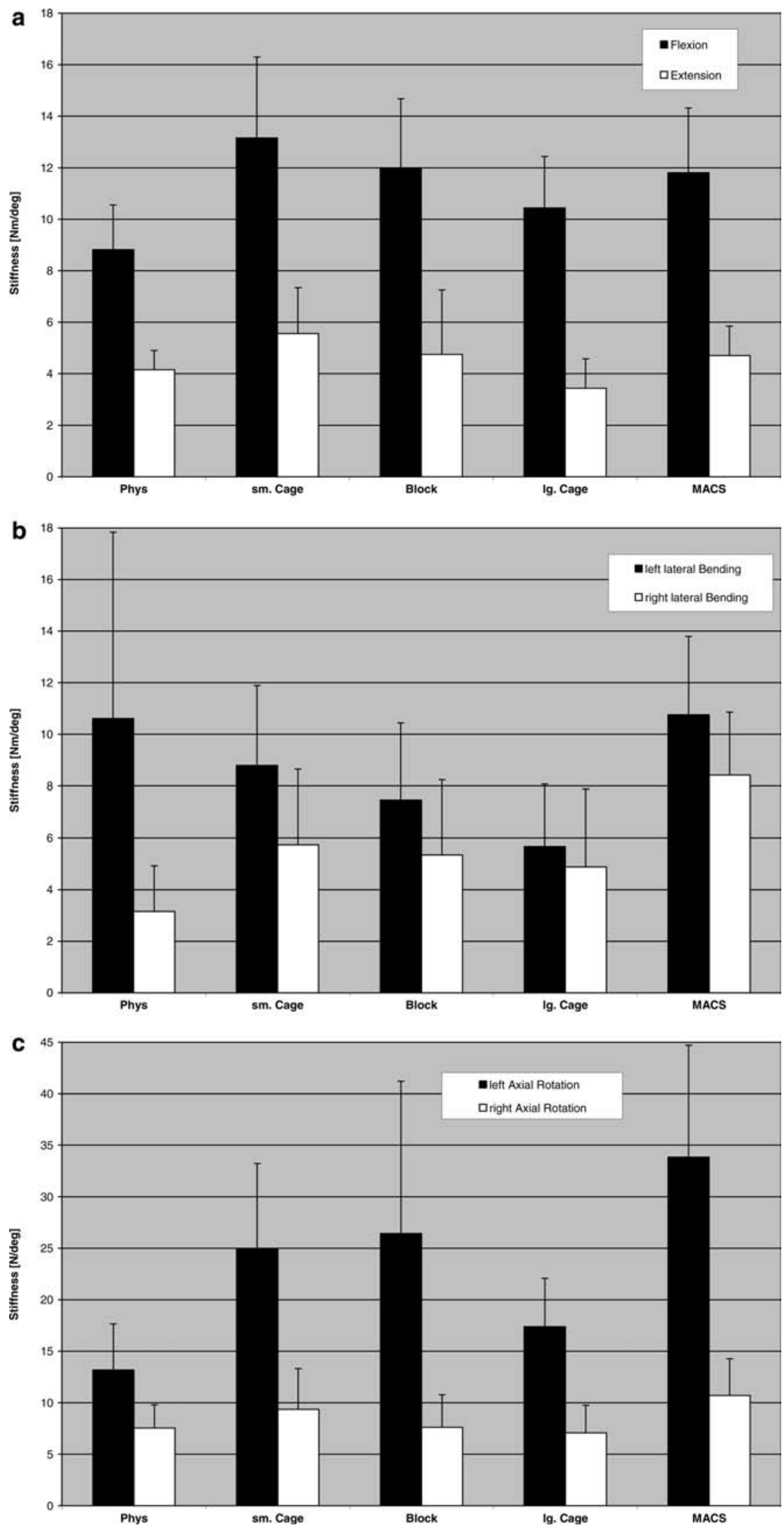
The NZ for the flexion/extension using the PMA of the five groups was also computed (Fig. 3). The NZ of the large cage for flexion/extension was observed to be significantly larger than the physiologic specimens ($p = 0.007$). Furthermore, in the instrumented groups a tendency towards smaller NZs compared to the physiologic specimens was observed. The NZ of the large cage was significantly larger than the NZ of the small cage ($p = 0.005$), the block ($p = 0.05$) and the MACS alone ($p = 0.021$) (Fig. 3). The NZ for lateral bending for physiologic specimens was significantly larger ($p \leq 0.001$) than in all the instrumented groups. No statistical significance between the instrumented groups was observed in lateral bending. Regarding the NZ for axial rotation, no statistical significance was observed between the physiologic specimens and the instrumented or between the individual instrumented groups, with the exception of the MACS-plate alone, which exhibited a significantly smaller NZ than the physiologic specimens (Fig. 3). In general, the NZ of the small cage was less in all directions tested than that of the block, which was in turn less than that of the large cage, although these differences where not always statistically significant (Fig. 3).

The bending stiffness (BES) in flexion with the PMA is summarized graphically (Fig. 4a). The small cage ($p < 0.001$), the block ($p < 0.001$), the large cage ($p = 0.042$) and the MACS-plate alone ($p < 0.001$) are statistically significantly stiffer than to the physiologic specimens. Furthermore, the small cage was significantly stiffer than the large cage ($p = 0.007$). In extension, bending stiffness of the small cage was significantly greater than the physiologic specimens ($p = 0.004$). Moreover, the stiffness of the small cage ($p = 0.001$) and the block ($p = 0.029$) were both significantly greater than that of the large cage. The left lateral bending stiffness of the large cage was similarly significantly smaller than the physiologic specimens ($p = 0.014$) as well as the MACS alone ($p = 0.045$). Further significant differences between groups were not observed. All instrumentations were significantly stiffer than the physiologic specimens for right lateral bending (small cage $p = 0.002$, block $p = 0.001$, large block $p = 0.041$, and MACS $p < 0.001$). Furthermore, the MACS-plate was statistically significantly stiffer than the small cage ($p = 0.0119$), the block ($p = 0.0041$), and the large cage ($p = 0.0010$).

In comparison to the physiologic specimens, the small cage ($p < 0.001$), the block ($p < 0.001$) and the MACS-plate ($p < 0.001$) were significantly stiffer for left rotational stiffness. The MACS-plate was also stiffer compared to the other instrumented specimens (small cage $p = 0.013$, block $p = 0.034$, large block $p < 0.001$). Finally, the small cage ($p = 0.036$) and the block ($p = 0.012$) were statistically significantly stiffer in comparison to the large block. Of the instrumented groups, only the MACS-plate was statistically significantly stiffer than the physiologic specimens ($p < 0.001$) for right rotational stiffness. It was also stiffer than the large cage ($p = 0.006$) and the block ($p = 0.018$). Further statistically significant differences were not observed, although the stiffness of the small cage and the block always tended to be greater than of the large cage.

Bending stiffness at an offset of 20 mm (Flex20) measured with the MTS is reported relative to the values

Fig. 4 **a** Stiffness for flexion and extension with the PMA. **b** Stiffness in lateral bending with the PMA. **c** Stiffness in axial rotation to the left and right with the PMA



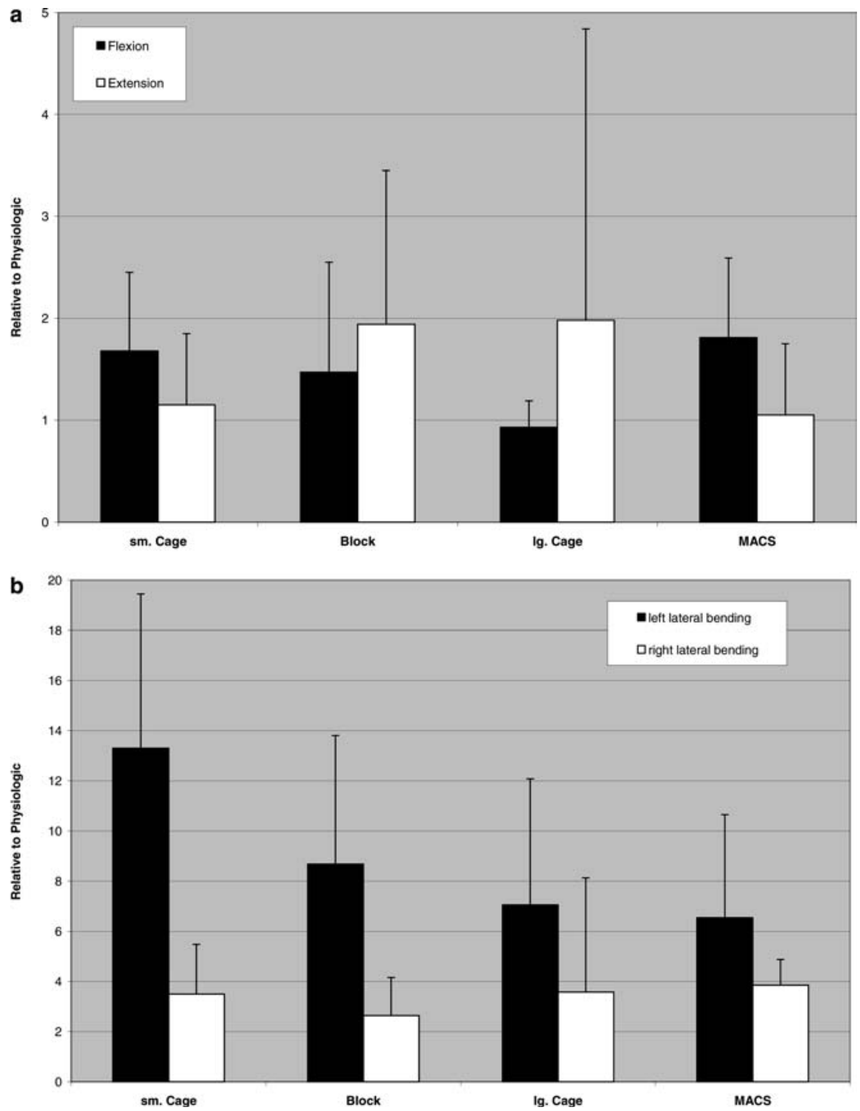


Fig. 5 **a** Stiffness in flexion and extension on the MTS with an offset of 20 mm. Values are given relative to the physiologic specimens. **b** Stiffness for lateral bending with an offset of 30 mm in the MTS. Values are given relative to physiologic

measured on the physiologic specimens (Fig. 5a). The small cage ($p = 0.040$) as well as the MACS-plate alone ($p = 0.016$) were significantly stiffer in comparison to the large cage. Bending stiffness in extension with an offset of 20 mm (Ext20, Fig. 5a) was observed to be greater for all instrumented groups in comparison to the original specimens (large cage 198%; block 194%; small cage 115%; MACS 105%). No statistically significant differences were observed between the individual instrumented groups.

In left lateral bending with an offset of 30 mm (see Fig. 5b), the small cage was significantly stiffer than the large cage ($p = 0.016$). Further differences were not observed. Similar values for right lateral bending with a 30 mm offset were observed. The relative rigidity of the four instrumentations was comparable (265–386% of physiologic stiffness). Statistical significances were not observed.

Discussion

Ten (10) bovine lumbar spine motion segments were used for this investigation. Generally, human motion segments are recommended for the biomechanical stability investigations of vertebral column implants since the pattern of mobility and the size of the segments in animals tend to differ from that of the human spine [7, 40, 58]. No problems with regards to implant size arose in the study presented through the use of calf lumbar spines in combination with instrumentation designed for human use. The optimal implant sizes were determined by in preliminary experiments. On the other hand, the availability of human vertebral column segments is limited due to ethical, religious and other reasons and are increasingly difficult to obtain because of legal restrictions [64]. For this reason, animal vertebral columns

were used for testing since these are readily available. The biomechanical tests were carried out using vertebral columns from pigs [38], goats [6, 45], sheep [63], rabbits [12, 22] and calves [1, 37, 67]. Nonetheless, calf lumbar spines appear to offer the best alternative as a substitute material since bone density is comparable to the human, and the pedicles are also of a similar size, although there is a complete growth plate in bovine animals as opposed to fused ring apophyses of human adults [10, 64]. Ideally the calves should be from 6 to 8 weeks old, unfortunately, calves at this age do not weigh enough to warrant slaughtering so that most investigations use animals of 6 months of age with a slaughter weight of about 170 kg [64]. The calves used for this study were between 5 and 7 months and weighed between 130 and 159 kg.

According to the recommendations of Wilke et al. [65], the preparations and tests were carried out at room temperature and the specimens packed into two vacuumed plastic bags and stored at −20°C. Panjabi had proven already in 1985 that the stability of specimens prepared in this way remained almost unchanged in comparison to unfrozen fresh specimens [44]. The total thawing time per segment remained consistently under 24 h. We avoided using a formaldehyde fixation since this altered the biomechanical properties of the specimens [65].

As early as 1988, Panjabi had defined decisive criteria for the biomechanical investigation of vertebral columns with the help of flexible testing systems [40], which Wilke adapted in his definition of test criteria of vertebral column implants [65]. The flexible testing system, the PMA, constructed with data from Panjabi [41] was used in the present study. The system complied with the above-mentioned stipulations of Wilke. Comparable flexible testing systems have been used by other authors for the investigation of biomechanical properties of motion segments [42, 43, 61, 65]. Additionally, an uniaxial testing system (MTS Minibionix 858) was used for further combined axial compression and bending loading. A similar system has also been used by numerous other authors [7, 17, 53, 56, 66].

The MACS TL-system, which supports mono-segmental or multi-segmental fusion of the lumbar vertebral column from a ventral aspect also with the endoscopic technique, was first used in 1994 and, is in the meantime being implemented increasingly more often [4, 19, 50–52]. This system is particularly suitable for comparative investigations of various interbody implants, such as the study presented herein, since the plates can be removed and manipulated in the intervertebral space without altering the primary position of the screws. After completion of the manipulation in the intervertebral space, the plate is re-fixed by applying a torque wrench to the screws. It was thus ensured that the stability of the plate osteosynthesis in all investigations was comparable with the initial condition and in this way the differing stabilities were influenced only by the interbody implants.

The Pyramesh cage was used as an interbody implant and has already been tested biomechanically in at least one study [21]. This cage is similar to the Harms cage, which has been used for numerous in-vitro investigations [25, 27, 28, 30, 33]. The height of the cage is cut from a longer section of a cage with the help of forceps. We used Biosorb as a bone substitute material comprised of tricalcium phosphate. The contact surface of the blocks with the bordering endplates is considerably larger than that of the cage. The porosity was ascertained at 30% and the maximal compressive strength 30 MPa. It is known from previous studies that ceramic material sometimes has tends to fracture, however, with no reported change in the position of the treated segment [9, 60]. It is also known from the literature that the average elasticity module of the lumbar cancellous bone of a 60-year old individual is about 25 MPa [46]. The required compressive strength of the bone substitute of 30 MPa was thus judged sufficient for the size of the blocks tested in order to avoid fractures during testing. In fact, the strength of the Biosorb blocks proved sufficient to withstand the loadings we applied and no fractures observed upon removal after testing. On some occasions, however, small avulsions were detected at the edges of the blocks but it is assumed that these had no biomechanical effect.

Titanium cages, such as the Harms cage or the Pyramesh cage, are used as a standard implants for interbody instrumentations of the lumbar vertebral columns. The latter was considered suitable for our investigation. Controversial statements have been made in the literature as to the way in which such an implant should be inserted into the intervertebral space. On the one hand, it is thought that the bordering endplates should be preserved since it is here that the greatest stability of the vertebral body is localized [21, 29, 62], and an impairment of the endplates would lead to shifting of the interbody implants [24]. For this reason, a number of authors conclude that the more of the endplates that is preserved, the more stable the implantation [13, 18, 21, 29, 31]. Other authors are of the opinion that the application of endplates for the stability of an implant is not so important, or that the implants will only shift if there is only a minimal contact surface between the implant and the end plate [8, 23, 37]. In order to enabling fusion, it is essential that the implant is connected to the blood system of the individual. Based on these studies one would conclude that the access to the blood supply required in order to assure disintegration would be counteractive to the high primary stability of the segment. However, not mentioned at all in the literature are comparable biomechanical investigations between ALIF with and without endplates using parameters (ROM, NZ, stiffness), which characterize the stability of the motion segment. A number of studies have made comparisons between the different cages in conjunction with ALIF and with the physiologic specimens [15, 25,

26, 37]. In 2000, Kanayama et al. undertook biomechanical investigations on calves using six different cages, one of which was the Harms cage, and observed no statistically significant differences [25]. Oxland published similar findings [37]. Lee reported a higher stability in the axial rotation of the Harms cage as opposed to the bone blocks in his biomechanical investigations [30]. In vivo investigations on goats gave suggested better results using the Harms cage as opposed to the bone blocks [6]. A biomechanical investigation with axial compression testing of the Pyramesh cage confirmed the results of other authors in relation to the size of the cages. The larger the inserted cage, the more stable it is [21].

Apart from substituting the intervertebral disk with cages filled with autologous spongiosa, the possibility of using bone substitute materials also exists. Available materials are organic substances such as coral products (Interpore, Biocoral), bovine substances (Endobon and Pyrost), and synthetic anorganic substances (calcium-biphosphate, tricalcium phosphate, hydroxylapatite, and composite of substances). The anorganic substances are favored on account of the immunologic and infectious problems associated with biological substances [20, 47, 48, 57]. Of the anorganic substances available, tricalcium phosphate is most frequently used since contrary to hydroxylapatite it can be absorbed and thus potentially completely absorbed by a bony fusion [20, 57]. For the biological integration of the ceramic material, both the substance and the structure are decisive factors. The larger the pores and the higher the porosity, the quicker the osteointegration. The opposite can be said for the stability of ceramic material, whereby the porosity of ceramic material should amount to at least 30% for osteointegration to occur [20, 47, 48, 57, 60]. Investigations of potential applications of ceramic materials entailed implantation into all the different regions of the spinal column in various animal species (hare, dog, sheep, and goat) [9, 11, 34, 35, 45]. A potential advantage of these blocks is that they can be produced in any desired size so as to optimize the contact surface between the endplates and the implant. The larger the contact surface between the implant and the endplates, the higher the primary stability [8, 24].

In our investigations the ROM was significantly reduced in comparison to the physiologic specimens for all three of the interbody implants (small Pyramesh cage with endplates, Biosorb-block and large Pyramesh cage without endplates) when tested with anterior MACS-plate for in all tested directions of movement. Oda et al. [36] performed investigations on human segments, also using the Harms cage and a ventral double rod system (Kaneda SR), with the MTS in flexion/extension and lateral bending up to 4 Nm. Compared to the physiologic motion segments, contrary to our findings, he could only determine a significant reduction in the ROM for the lateral bending but not for flexion/extension [36]. Knop et al. also published a biomechanical test of the totalcorporectomy model in 2001 on human specimens, whereby they compared the Harms cage and an expandable vertebral body replacement (Synex) in combination with an anterior fixator (Ventrofix) [27]. In flexion/extension ROM was found to be increased compared to physiologic, in lateral bending diminished, however not significantly, while in rotation both cages resulted in significant increase in ROM. For these very varying results, either the greater stability of the MACS-plate was responsible, as opposed to the model used by Knop et al., or the corporectomy in that model was responsible, which could of course result in greater instability compared to the simple intervertebral disk substitute investigated herein. Our study allowed us to make a comparison of the three very different interbody implants. The ROM of the large cage always tended to be the largest of the three implants, and with the exception of axial rotation significantly so.

Furthermore, the NZ of the large cage was always larger than that of the small cage and the block: for flexion/extension this was again significant in comparison with both. In contrast to the physiologic specimens, the NZ always tended to be smaller for the block and the small cage but only significantly in lateral bending. The large cage, on the other hand, demonstrated a generally larger NZ than physiologic, significantly so in flexion. In the case of Knop et al. [27], the NZ for both tested implants for vertebral body replacement for flexion/extension and rotation was significantly larger than with the physiologic specimens. The difference was insignificant with lateral bending. Thus, similar to the results for ROM, clear differences exist for the NZ of the implants in comparison with our data.

From the results of the stability investigations using the PMA it can be seen that the small cage with retained endplates and the block without endplates are statistically stiffer than physiologic in all directions with the exception of left lateral bending. On the other hand, the large cage is usually less stable than physiologic and both the other cages. The situation with MTS serves to prove another point. Under large axial loading all three implants are more rigid compared to the physiologic specimens with the exception of flexion for the large block. Significant differences between the three implants were revealed only for flexion and left lateral bending, here the small cage showed an advantage over the large cage.

Removal of the endplates (large cage) resulted in a significant loss of primary stability as opposed to preserving the endplates (small cage). However, our results suggest that this problem can be avoided by the implantation of a ceramic block, therefore considerably enlarging the contact surface. For general clinical purposes, this would suggest that when using a cage, preservation of the bordering endplates should be sought with

an anterior stabilization, otherwise the primary stability may be inadequate. If a reliable biological osteointegration is to be achieved with removal of endplates, then a ceramic block with a considerably larger contact surface should be inserted in order to ensure sufficient primary stability. To what extent this ceramic material with the parameters tested in this study of (approximate failure strength 30 MPa, height. 1.5 cm, and porosity 30%) can achieve biological stability of the lumbar vertebral column remains to be shown. Such questions as well as the possible influence of the addition of cytokine such as BMP, must still to be clarified, for example by means of an in vivo animal model.

Conclusion

The removal of the bordering endplates leads to a significant reduction in primary stability during ALIF augmented with the MACS plate in a lumbar calf model. The primary stability of ALIF with a Biosorb block without endplates is comparable to that of a cage with retained endplates. The primary stability of ALIF with the cage and retained endplates and the block without endplates is significantly higher than the stability of the physiologic specimens, this is not the case for the cage upon removal of the endplates.

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