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ORIGINAL ARTICLE

Biomechanical advantage of C1 pedicle screws over C1 lateral mass screws: a cadaveric study

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Abstract
Purpose The established technique for posterior C1 screw placement is via the lateral mass. Use of C1 monocortical pedicle screws is an emerging technique which utilizes the bone of the posterior arch while avoiding the paravertebral venous plexus and the C2 nerve root. This study compared the relative biomechanical fixation strengths of C1 pedicle screws with C1 lateral mass screws.
Methods Nine human C1 vertebrae were instrumented with one lateral mass screw and one pedicle screw. The specimens were subjected to sinusoidal, cyclic (0.5 Hz) fatigue loading. Peak compressive and tensile forces started from ±25 N and constantly increased by 0.05 N every cycle. Testing was stopped at 5 mm displacement. Cycles to failure, displacement, and initial and end stiffness were measured. Finally, CT scans were taken and the removal torque measured.
Results The pedicle screw technique consistently and significantly outperformed the lateral mass technique in cycles to failure (1,083 ± 166 vs. 689 ± 240 cycles), initial stiffness (24.6 ± 3.9 vs. 19.9 ± 3.2 N/mm), end stiffness (16.6 ± 2.7 vs. 11.6 ± 3.6 N/mm) and removal torque (0.70 ± 0.78 vs. 0.13 ± 0.09 N m). Only 33 % of pedicle screws were loose after testing compared to 100 % of lateral mass screws.
Conclusions C1 pedicle screws were able to withstand higher toggle forces than lateral mass screws while maintaining a higher stiffness throughout and after testing. From a biomechanical point of view, the clinical use of pedicle screws in C1 is a promising alternative to lateral mass screws.

Keywords Atlas · Biomechanical fixation strength · Pedicle screw · Lateral mass screw · Stiffness

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Introduction

Instabilities of the upper cervical spine in the occipitocervical region may result from traumatic injuries, tumors, infections, rheumatoid arthritis or congenital anomalies. Different stabilization techniques have been described in the literature and are established for clinical use to avoid persistent instability and neurological complications [1–4]. Unfortunately, due to the proximity of neurovascular structures, these operative techniques are technically challenging and associated with a significant complication rate [5–8].

In 2001, Harms and Melcher [4] published a technique involving posterior C1–C2 fusion using polyaxial screws and rod fixation. Using this technique the placement of the screws in C1 is realized via the lateral mass. The entry point for the lateral mass screw is covered by a paravertebral venous plexus which if pierced can lead to extended bleeding and the C2 nerve root can be injured intraoperatively [7, 8]. To increase the mechanical stability of lateral mass screws, bicortical placement can be used [8, 9]. However, this endangers the hypoglossal nerve and the internal carotid artery [10–12].

Monocortical C1 pedicle screws are an emerging technique in which the screws are implanted via the posterior arch rather than the lateral mass. Clinically, this placement requires a careful subperiosteal preparation of the posterior aspect including the exposure of the vertebral artery at the superior rim [7, 13]. Therefore, the exposure of the lateral mass and accompanying structures, including the venous plexus and the C2 nerve root, is not necessary. In first clinical studies pedicle screw instrumentation was found to be effective and safe by leading to a solid arthrodesis with no signs of neurovascular complications or implant failure [7, 14, 15].

Biomechanically, the pedicle screw trajectory passes through the posterior arch enabling the use of longer screws with a greater bone contact area than lateral mass screws. A longer bone–screw interface has been shown to lead to improved screw stability [16]. Several biomechanical studies have investigated lateral mass screw–rod constructs [9, 17–21]. Recently, the pullout strengths and biomechanical stabilities afforded by C1 lateral mass screws and C1 pedicle screws using bicortical and monocortical fixation techniques have been analyzed by Ma et al. [8]. They showed similar pullout resistance and three-dimensional stability for bicortical C1 lateral mass screws and monocortical C1 pedicle screws; unicortical lateral mass screws provided the lowest pullout strength.

To our knowledge, a biomechanical comparison of monocortical C1 pedicle screws with monocortical C1 lateral mass screws has not yet been described for cranial–caudal toggling, which is considered to be more physiological than axial pullout of the screw. Therefore, the aim of this ex vivo study was to compare the relative biomechanical fixation strengths of monocortical C1 pedicle screws versus monocortical C1 lateral mass screws in terms of cycles to failure, stiffness and removal torque by applying a cranial–caudal toggling force directly at the screw head.

Materials and methods

Specimens

After approval by the local ethics committee, nine human cadaveric C1 vertebrae (58.0 ± 11.1 years, three female, six male) were received from the local institute of forensic medicine. Specimens were sealed in plastic bags and stored at −20 °C.

Preoperative CT scans (0.90 mm slice thickness, 0.45 mm spacing, Mx8000 IDT 16, Philips Healthcare, DA Best, NL) were performed with a phantom (QRM-BDC, QRM, Möhrendorf, DE) and were used to exclude pathological malformations or preexisting fractures as well as to determine the apparent volumetric bone mineral density (BMD). Trabecular volumetric BMD was determined by segmenting a 25 × 25 × 25 voxel cube from the center of each lateral mass (Avizo version 5.1, Mercury Computer Systems, San Diego, CA, USA). The average Hounsfield unit value was scaled linearly to the reference densities of the phantom.

Preparation, fixation and instrumentation

The night before testing, the specimen was thawed to room temperature. On the day of testing all soft tissue was removed. After sample preparation, a custom-made mold was created for each atlas (Fig. 1). To imitate physiologic boundary conditions, the mold only contacted the cranial and caudal surfaces of the facet joints. The mold was created by first placing modeling clay through the center of the atlas, over the transverse processes and along the anterior arch (Fig. 1a). The clay-covered atlas was then aligned in a rectangular base mold (Fig. 1b). Finally, both sides of the mold were cast at the same time by filling the mold (posterior arch facing up) with a polyurethane resin (Ureol FC53, Gößl & Paff, Karlskron, DE) to produce level, parallel endplates for mounting. After hardening, the mold was taken apart and the clay removed (Fig. 1c).

All vertebrae were bilaterally instrumented with polyaxial screws of the same size (outer diameter 3.5 mm, length 26 mm, Synapse System, Synthes GmbH, Oberdorf, CH; Table 1). Each vertebra received one polyaxial lateral mass screw (cortical profile, 10 mm unthreaded shaft) and one polyaxial pedicle screw (cancellous profile). The side on which each screw was placed was equally allocated based on BMD, age, gender and testing order (Table 1).

The entry point differed for the two techniques (Fig. 2). The lateral mass screw was inserted at the crossing of the inferior rim of the posterior arch and the middle of the lateral mass, aiming at the center with a cephalad angulation of 20°. The pedicle screw entered through the posterior arch, aiming at the center of the lateral mass, without perforating the superior rim of the lamina. Thus, the screw tips converged to the same height and depth within the bone without touching the anterior cortex (Fig. 2). To ensure consistency, all instrumentation was performed by the same orthopedic surgeon and postoperative X-rays were taken in the axial, antero-posterior and lateral planes to confirm proper implant positioning.

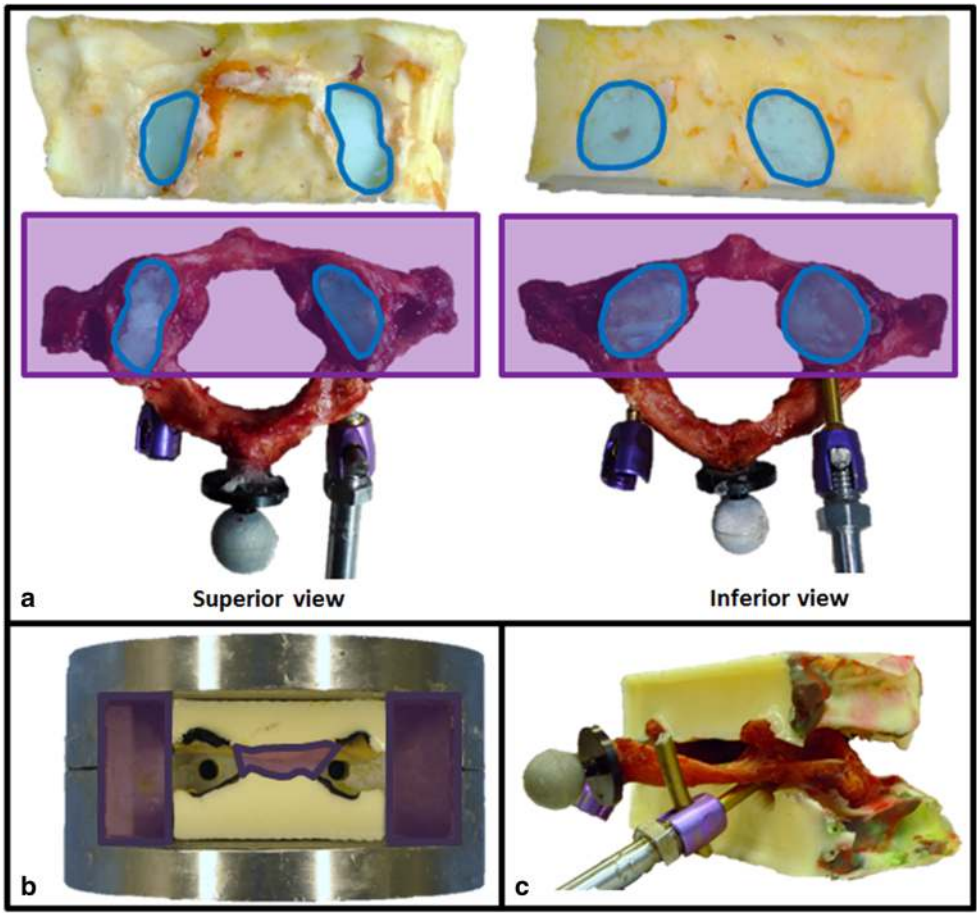

Fig. 1 Creation of a custom-made mold for each atlas. **a** An example of a mold highlighting the contact area (*blue*) which only occurred at the cranial and caudal surfaces of the facet joints. **b** Rectangular base mold in which the molds were cast. **c** 3D perspective of a mold with an instrumented specimen. *Purple* highlights (**a**, **b**) show where clay would be placed

Table 1 Testing group characteristics showing properties for both the screw design and the specimen characteristics. Since each vertebra was allocated one lateral mass and one pedicle screw, the number, age and gender balance of the groups were equal

	Screw design		Specimen characteristics
Massa screw ($N = 9$)		Outer diameter (mm) = 3.5 Core diameter (mm) = 2.95 Thread length (mm) = 16 Thread pitch = 1.25	Age (years) = 58.0 ± 11.4 BMD (gHA/mm^3) = 210.5 ± 71.6 Gender = three females, six males
Pedicle screw ($N = 9$)		Outer diameter (mm) = 3.5 Core diameter (mm) = 2.95 Thread length (mm) = 26 Thread pitch = 1.75	Age (years) = 58.0 ± 11.4 BMD (gHA/mm^3) = 209.8 ± 71.5 Gender = three females, six males

Loading modalities

After instrumentation, the specimens were clamped to a servohydraulic testing machine (858 Bionix®, MTS, Eden Prairie, MN, USA, Fig. 3). A sinusoidal, cyclic (0.5 Hz) force was applied to the screw head, which was free to rotate around the transverse axis with all other degrees of freedom (DOF) fixed. The screw head was attached to the testing machine with a customized steel locking adaptor and a 15 mm long, 3.5 mm diameter rod, to lock the polyaxial head in place. Cranio-caudal peak compressive and tensile forces started from ±25 N and increased linearly by 0.05 N every cycle. Toggle testing was stopped when 5 mm displacement was achieved caudally. After

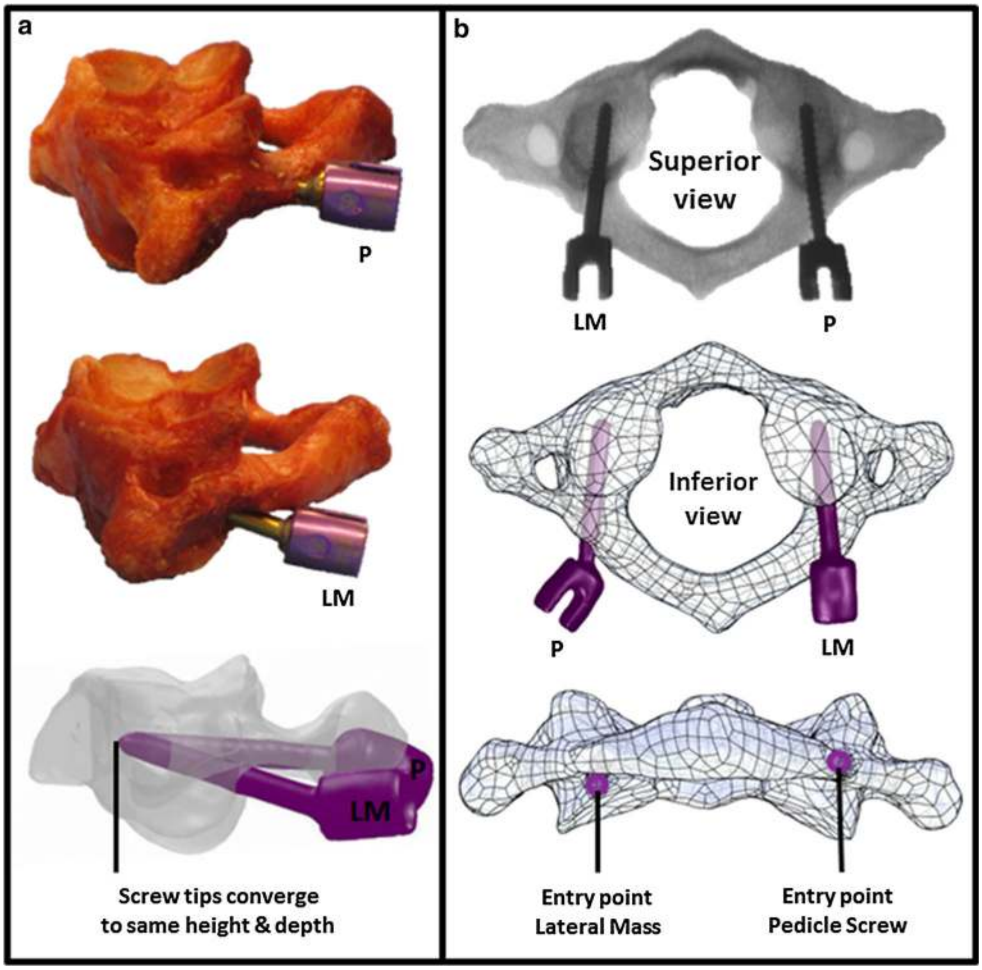

Fig. 2 Instrumentation of C1 with one lateral mass screw (LM) and one pedicle screw (P). **a** Views showing that the cephalad angulations (*top* and *middle*), depth and height match (*bottom*). **b** X-ray (*top*) and rendered CT scans showing the relative depths (*middle*) and entry points (*bottom*) of each technique

fatigue failure of one screw, the contralateral screw was loaded with the same protocol. Testing conditions were at room temperature and specimens were sprayed with Ringer solution during preparation and before testing to maintain tissue hydration. Finally, CT scans were taken with the screws in situ and the maximal removal torque was then measured (Torsiometer 760, Stahlwille, Wuppertal, DE).

Data analysis

Cycles to failure, displacement, removal torque, and initial and end stiffnesses were output variables. Stiffness was defined by the slope of the caudal force–displacement loading curve. The slope was calculated from a linear best fit for each cycle with the first and last 10 % of the cycle's data not included to ensure only the linear portion was considered. Initial stiffness was defined from the first caudal loading cycle and end stiffness from the last. The screw length into the vertebra and the cephalad angle was measured directly from the post-testing CT scans (Avizo version 5.1).

Statistical evaluations were performed with the software package SPSS Statistics 20 (IBM Corp., Armonk, NY, USA) with a type I error probability set to 5 %. A paired-samples *t* test was used to test differences between groups (lateral mass vs. pedicle screws) and the effect size between the groups was determined with a point-biserial correlation. The assumption that the sampling distribution of the differences is normally distributed was upheld for the variables: cycles to failure, force at failure, initial stiffness, end stiffness and removal torque. A Pearson correlation coefficient was used to measure covariance. Effect size magnitude is represented by adjusted R^2 values to account for small sample sizes and was determined by linear regression.

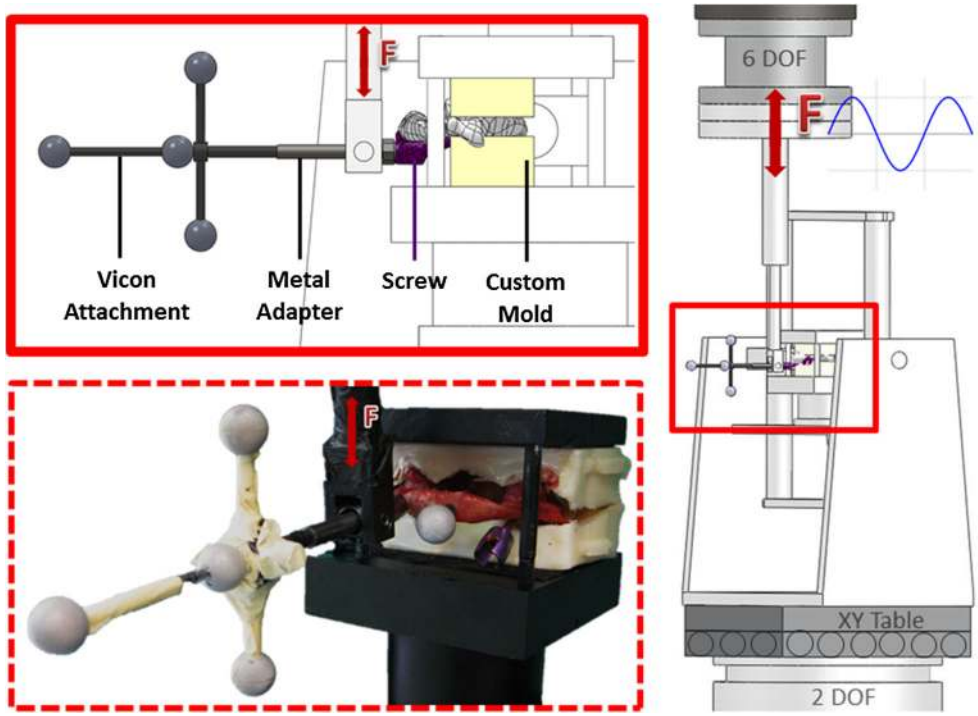

Fig. 3 Schematic of mechanical test setup with *red box* denoting an expanded view of the clamped specimen attached via the custom-made metal adapter. The *red dashed box* is a photograph of the clamped specimen

Table 2 Output parameters with the averages and standard deviations given by group

	Massa screw	Pedicle screw	Significance
Cycles to failure (#)	689 ± 240	1083 ± 166	$**p = 0.004, R^2 = 0.48$
Force at failure (N)	−58.3 ± 13.3	−76.9 ± 11.0	$**p = 0.008, R^2 = 0.36$
Initial stiffness (N/mm)	19.9 ± 3.2	24.6 ± 3.9	$**p = 0.002, R^2 = 0.29$
End stiffness (N/mm)	11.6 ± 3.6	16.6 ± 2.7	$**p = 0.008, R^2 = 0.18$
Removal torque (N m)	0.13 ± 0.09	0.70 ± 0.78	$*p = 0.04, R^2 = 0.18$
Screw depth (mm)	15.4 ± 0.6	21.9 ± 1.0	$**p < 0.001, R^2 = 0.94$
Cephalad angle (°)	23.6 ± 5.5	10.2 ± 3.1	$**p < 0.001, R^2 = 0.70$
Loose (torque)[a]	100 %	33 %	

[a] Clinical loosening is defined as a removal torque of less than 0.40 N m [25]

Results

Fixation strength

The pedicle screw technique consistently and significantly outperformed the lateral mass technique in all measures of bone–screw fixation strength analyzed (Table 2). This included cycles to failure ($p = 0.004$, $R^2 = 0.48$), force at failure ($p = 0.008$, $R^2 = 0.36$), initial stiffness ($p = 0.002$, $R^2 = 0.29$), end stiffness ($p = 0.008$, $R^2 = 0.18$) and removal torque ($p = 0.04$, $R^2 = 0.18$). Distinct differences were seen in the progression of both displacement and stiffness between groups (Figs. 4, 5). The screw loosening rate was higher for the lateral mass screws (100 %) than for pedicle screws (33 %).

Surgical technique effects

The screw depth and the cephalad angle were different between surgical techniques ($p < 0.001$) and correlated negatively with each other ($p < 0.001$). With increasing screw depth, there were increases in cycles to failure ($p < 0.001$, $R^2 = 0.57$), compressive force ($p = 0.003$, $R^2 = 0.39$), initial stiffness ($p = 0.003$, $R^2 = 0.40$), end stiffness ($p = 0.001$, $R^2 = 0.45$) and removal torque ($p = 0.04$, $R^2 = 0.19$). With increasing cephalad angle, cycles to failure ($p = 0.001$, $R^2 = 0.47$), compressive force ($p = 0.02$, $R^2 = 0.25$), initial stiffness ($p = 0.01$, $R^2 = 0.30$) and end stiffness ($p = 0.01$, $R^2 = 0.29$) all decreased. As desired, the sagittal angle did not vary with the surgical technique ($p = 0.19$).

Specimen geometry and bone quality

The specimens were of normal bone quality (BMD = 210 ± 71.6 mgHA/cm^3). BMD did not correlate significantly with any measured parameters, including cycles to failure, force to failure, removal torque, age, gender or stiffness values ($p > 0.14$). Specimen width correlated positively with age ($p = 0.008$, $R^2 = 0.32$) and was higher for male specimens ($p = 0.02$); specimen height correlated positively with initial stiffness ($p = 0.01$, $R^2 = 0.30$) but not with end stiffness ($p = 0.10$).

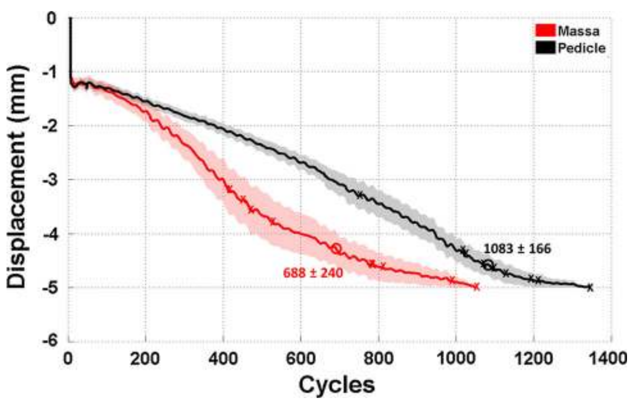

Fig. 4 Graph of the average displacement vs. cycle number, showing pedicle screws having a much higher resistance to structural deformation. After about 100 cycles, a diverging of the displacement curves exists. An '×' represents the cycle number for a specimen failure, *circles* are the averages and the *shaded areas* represent the standard error

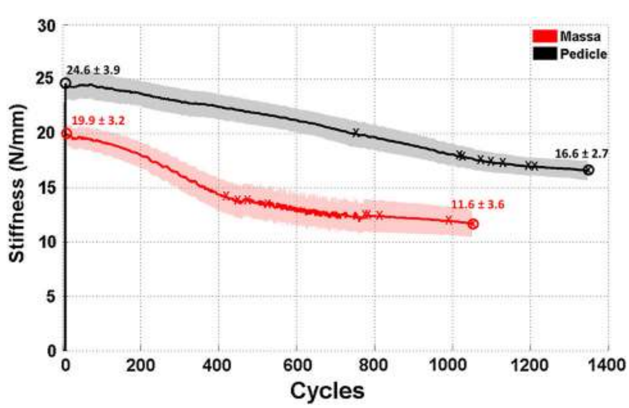

Fig. 5 Graph of the average stiffness vs. cycle number, showing pedicle screws having a significantly higher initial and end stiffness. Stiffness loss (distance between curves) was similar throughout testing. An '×' along the curve represents a cycle number where a specimen failed and the *shaded areas* represent the standard error

Discussion

The placement of monocortical C1 pedicle screws is an emerging technique which enables the use of longer screws with a greater bone contact area in comparison with monocortical C1 lateral mass screws [22]. While C1 bicortical lateral mass screws may also have superior biomechanical stability compared to monocortical C1 lateral mass screws, they also bear the risk of rare but extremely serious clinical complications such as injuries to the carotid artery or the hypoglossal nerve [10–12]. Therefore, the biomechanical properties of monocortical C1 lateral mass screws were compared to C1 pedicle screws.

In our study, the biomechanical superiority of C1 pedicle screws was proven over C1 lateral mass screws with regard to cycles to failure, stiffness, removal torque and loosening criteria ($p < 0.04$; Table 2). In fact, the longest lasting lateral mass screw (1,048 cycles) did not reach the average cycles to failure of the pedicle screw group (1,083 ± 168 cycles; Fig. 4). Under repeated loading, pedicle screw placement provides a higher resistance to structural deformation, shown by the diverging average displacement curves after 100 cycles (Fig. 4). The pedicle screw technique also started with, and maintained, a higher stiffness during testing (Fig. 5), showing that pedicle screw placement provides primary stabilization as well as longer-term fixation benefits. These findings correspond well with the results of a recent study where unicortical C1 lateral mass screws showed weaker pullout strength in comparison with C1 pedicle screws [8].

In accordance with previous findings, an increased screw depth in bone has been shown to improve fixation strength in vertebrae [16]. The 6.5 mm (42 %) greater depth for the pedicle projection compared to lateral mass projection is in line with a previous anatomic study which demonstrated the mean intraosseous depth to be 10 mm higher [22].

There was an inverse correlation of cephalad angle and screw depth ($p < 0.001$). This is expected because a higher cephalad angle was necessary for the tip of the lateral mass screw to achieve the same implantation height as the pedicle screw in the sagittal plane due to a more inferior entry point. Less depth was achieved since the bone stock of the posterior arch was not utilized.

The current toggling fatigue setup was utilized because it allows for the progressive measurement of loosening parameters over time and within the physiologic range. Furthermore, it produces an in vivo-like failure pattern with loosening zone expansion at the screw tip. In contrast, pullout testing produces purely primary stability data (no effect of time) for a non-physiologic force. Loading applied directly to the screw head eliminates factors related to the screw–rod connections. A repeated, continually increasing force was used to achieve a sweep of forces within the physiological range, as well as an accelerated failure to reduce the effects of specimen degradation.

In recent biomechanical studies, different posterior stabilization techniques have been investigated for the atlantoaxial complex [9, 17–21]. Superior stability was demonstrated for transarticular screws, which pass through the C1–C2 articulation, in comparison with established posterior wiring techniques [17, 18]. Unfortunately, in up to 23 % of the cases, safe placement of these transarticular screws is not possible due to anatomic limitations of the vertebral artery [13, 14, 23].

C1 lateral mass screw–rod constructs were established as an alternative technique with greater usability when compared with transarticular screws because they have shown similar stability and fewer anatomical constraints [18–21]. C1 screws combined with separate instrumentation of C2 allow for the active manipulation and reduction in non-mobile atlantoaxial (sub-) luxation which is not possible with transarticular screws [24]. The monocortical pedicle screw technique has potential unevaluated clinical benefits over the lateral mass technique, including a minimization of blood loss and a decreased neurovascular complication rate [7, 8, 10–12].

Anatomical studies have demonstrated the potential of inserting a 3.5 mm C1 pedicle screw safely through the posterior arch [14, 15, 22, 23]. Nevertheless, the thinnest height of the screw tract underneath the groove of the vertebral artery (to which close intraoperative attention must be paid) was measured to be less than 4 mm in 8–31.7 % of specimens analyzed. Therefore, preoperative CT scans are necessary to determine the anatomic dimensions, plan acceptable screw projection and to evaluate the feasibility of C1 pedicle screw placement. These are necessary to avoid "over-dimensioned" screws which may result in the splitting of the posterior arch and, therefore, compromise the rate of bony fusion. Unacceptable screw projection may occur in up to one-third of all cases, especially in patients with rheumatoid arthritis. For these cases, the placement of C1 lateral mass screws is the more appropriate alternative.

The limitations of this ex vivo study include the natural variation of the specimens, small sample sizes and the variation of the pitches for the screw threads. Natural variation was reduced as far as possible by utilizing a repeated measures design: with one lateral mass screw and one pedicle screw per specimen.

In summary, the C1 pedicle screw technique appears to be a promising alternative to the C1 lateral mass screws due to improved biomechanical stability. The fixation advantages of using monocortical C1 pedicle screws probably arises from a greater insertion depth into the bone, as well as a smaller canal width which may increase compression on the screw threads. From a clinical point of view, this procedure offers a potential to minimize blood loss and to decrease neurovascular complications which suggest both an economic and a clinical improvement. Nevertheless, as in all atlantoaxial stabilization procedures, careful attention must be paid to ensure the proper spacing of the screw tract from the vertebral artery and the nerves because of the potential catastrophic consequences of a misplacement. Further clinical studies are necessary to evaluate this procedure in appropriate detail.

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Conflict of interest Nils Hansen-Algenstaedt is working as a consultant for Stryker, DepuySynthes and SpineArt. He receives Royalties on implants from Stryker and Globus Medical. No other authors have any declarations.

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