Evaluation of biomechanical properties of anterior atlantoaxial transarticular locking plate system using three-dimensional finite element analysis

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Abstract

Purpose To evaluate a new anterior atlantoaxial transarticular locking plate system using finite element analysis.

Methods Thin-section spiral computed tomography was performed from occiput to C₂ region. A finite element model of an unstable atlantoaxial joint, treated with an anterior atlantoaxial transarticular locking plate system, was compared with the simple anterior atlantoaxial transarticular screw system. Flexion, extension, lateral bending, and axial rotation were imposed on the model. Displacement of the atlantoaxial transarticular screw and stress at the screw–bone interface were observed for the two internal fixation systems.

Results Screw displacement was less using the anterior atlantoaxial transarticular locking plate system compared to simple anterior atlantoaxial transarticular screw fixation under various conditions, and stability increased especially during flexion and extension.

Conclusions The anterior atlantoaxial transarticular locking plate system not only provided stronger fixation, but also decreased screw-bearing stress and screw–bone interface stress compared to simple anterior atlantoaxial transarticular screw fixation.

Keywords Atlantoaxial fusion · Anterolateral retropharyngeal approach · Biomechanics · Finite element analysis

Introduction

Atlantoaxial instability is often treated with posterior atlantoaxial short-segment fusion techniques, including posterior wire internal fixation [1], posterior transarticular screw [2], lamina clamp (Halifa clamp or Apofix), atlantoaxial lateral mass screws, and pedicle screws [3, 4]. Although these surgical procedures retain atlanto-occipital joint motion and other cervical segment motions, complications such as internal fixation breakage and loosening are likely to occur. Stillerman and Wilson [5] and Coyne et al. [6] achieved satisfactory clinical outcomes in the treatment of atlantoaxial instability using combined transarticular screw fixation and Brooks cable fixation. However, this technique requires anatomical reduction of the atlantoaxial joint. Harms and Melcher [3] treated atlantoaxial instability using the posterior atlas lateral mass screw/axis pedicle screw technique, where the angle of screw insertion was small and anatomical reduction of the atlantoaxial joint was not required. Biomechanical evaluation showed that its mechanical properties were similar to those of the posterior transarticular screw fixation technique [7].

However, since posterior internal fixation is difficult to perform in some patients with bone defects in the posterior arch of the atlas, and challenging in atlantoaxial reduction or failed posterior surgery cases, anterior surgery becomes a viable alternative in these patients. At present, the most commonly used anterior fixation methods include the atlantoaxial transarticular screw system applied via

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anterolateral retropharyngeal approach and the screw–plate system applied via transoral approach.

In Schmelzle and Harms [8] introduced the concept of transoropharyngeal atlantoaxial plate fixation, i.e., the “Harms plate”. Kandziora produced an anterior atlantoaxial locking plate (AALP) based on the Harms plate and further improved AALP to produce the subarticular anterior atlantoaxial locking plate (SAALP). In China, Yin et al. [9] developed transoral atlantoaxial reduction plate (TARP). Because the Harms plate is an unlocking plate and the screws do not have locking function, complications arise such as screw loosening and insufficient stability. The SAALP plate has better stability and fixation strength [10]. TARP also has excellent stability and there is no significant difference between TARP and the combined Magerl screw and Brooks cable fixation technique. The combined Magerl screw and Brooks cable fixation technique is better than the single application of the Brooks technique, the Magerl technique, and anterior transaxial atlas lateral mass screw fixation. Although the transoral fixation is more precise, its surgical trauma is massive and many serious intra- and postoperative complications such as the cerebrospinal fluid leakage and wound infection may occur, limiting its clinical application [11–13].

Atlantoaxial facet screw fixation via anterolateral retropharyngeal approach is an ideal anterior surgical procedure with minimal soft tissue injury and relatively low risk of vertebral artery injury. It can also directly release the atlantoaxial joint and allow for removal of the articular cartilage [7, 14, 15] to prepare for spinal fusion. Emergence of the anterolateral retropharyngeal approach together with the use of endoscopic technique in cervical spine disease has greatly enhanced the clinical application of the anterolateral approach. Cases with irreducible atlantoaxial dislocation can be released and achieve satisfactory reduction through the anterolateral retropharyngeal approach under the assistance of the endoscope [16, 17]. However, this technique requires high quality anterolateral bone for the axis, and the stability is poor in flexion and extension. In addition, its biomechanical properties require further improvement [18].

At present, studies about the biomechanics of the cervical internal fixation devices are carried out mainly using cadaver experiments and finite element analysis. Among them, the cadaver experiment is mostly confined to the range of the three-dimensional movement of the spine and the internal fixation devices, and the pull-out strength of the screw. Finite element analysis simulates and analyzes various structures in the human body and their pathological changes using a computer. Its results are not affected by other factors and it can analyze the internal stress and strain, which are difficult to study using general experimental methods. Moreover, it has advantages such as the high accuracy and repeatability [19].

We used finite element analysis to evaluate treatment of atlantoaxial instability via the anterolateral retropharyngeal approach using an anterior atlantoaxial transarticular locking plate system. This system was based on the anterior transarticular screw fixation technique.

Methods

Design of anterior atlantoaxial transarticular locking plate system

The anterior atlantoaxial transarticular locking plate system includes a titanium plate, several screws, an atlantoaxial reduction device and other supporting surgical instruments (Fig. 1). The plate is composed of medical titanium alloy (Ti6Al4V). There are two oval-shaped locking screw fixation holes in the upper part of the plate used for inserting screws through the atlantoaxial joint. In addition, there are two round locking screw fixation holes in the lower part of the plate used for inserting screws into the axis body. A round hole is located at the center of the plate for reduction. During surgery, a screw can be placed temporarily in this hole and a reduction clamp can be inserted between the temporary screw and the anterior arch of the atlas. Atlantoaxial reduction can be achieved by distracting the clamp. The plate is 2 mm in thickness and internal catches are designed to lock four screws.

Fig. 1 Design of anterior atlantoaxial transarticular locking plate system

Finite element model of the upper cervical spine

A 21-year-old healthy male volunteer 171 cm tall and weighing 61 kg was selected as our cervical spine model after open mouth and lateral cervical spine X-rays excluded any congenital cervical spinal malformation or lesions. Spiral CT scanning was performed from the base of the occipital bone to C₃ vertebrae using 1 mm thick slices and the scanning data were collected and stored in Dicom format for the reconstruction of 3D bone structure using Mimics 13.0 software (Materialise, Leuven, Belgium). The geometric model was imported into Freeform Plus software (Geomagic Sensable group, Wilmington, MA, USA) for sanding, filling and denoising to form a geometric solid model which approximated a normal human cervical spine.

The geometric solid model was imported into Ansys 10.0 software (ANSYS, Inc. Canonsburg, PA, USA) for meshing and ligament loading according to start and end points of ligaments and cross-sectional area. This model consisted of 55,371 elements and 86,050 nodes. The model data are summarized in Table 1. The values in Table 1 were taken from a paper by Zhang et al. [21] which were derived from the work by Ng and Teo [20]. Based on the parameters of the model materials (Table 1), bone structures (including facet) and transverse ligaments were simulated using the eight-node tetrahedral elements. Other ligamentous structures (alar ligament, odontoid ligaments, nuchal ligaments, anterior longitudinal ligament, posterior longitudinal ligament, anterior and posterior atlanto-occipital membranes, ligaments of atlanto-occipital joint capsule and atlantoaxial joint capsule, interspinous ligament, supraspinous ligament) were simulated using a two-node cable element. At the articular surface of the lateral atlantoaxial joint, the generated thickness of the dens and transverse ligament both consisted of 0.5 mm articular cartilage. They possessed an elastic modulus of 10.4, Poisson's ratio of 0.4, and an articular surface coefficient of friction of 0.1. As both the atlantoaxial facet joints and their capsular ligaments were simulated, we also simulated the lower joint space. The definition of screw–bone interface in our model was close contact that ignored the micromotion between the screw and bone.

The range of motion at the base of the C₂ vertebra was defined in all directions as 0. Forty Newtons (N) of vertical downward pressure were imposed on the surface of the occipital condyle to simulate the weight of the head due to gravity. Approximately 1.5 Nm torque was imposed on the model from various directions to produce flexion, extension, lateral bending, and axial rotation. The range of motion at the atlantoaxial joint was calculated and compared with the results of in vitro tests performed by Panjabi [22] and Panjabi et al. [23]. The results from this comparison are shown in Table 2.

Table 1 Material properties used for various components of the model

Material	Young's modulus (Mpa)	Poisson's ratio
Bone ^a		
Cortical bone	12,000	0.29
Cancellous bone	450	0.29
Endplates	500	0.4
Posterior element	3,500	0.29
Annulus	3.4	0.4
Nucleus	1	0.49
Ligaments ^a		
ALL	30	0.3
PLL	20	0.3
ISL, LF	10	0.3
CL (C ₀ –C ₁)	1	0.3
CL (C ₁ –C ₂)	10	0.3
AIL	5	0.3
TL	20	0.3
ApL	20	0.3
NL	20	0.3

ALL anterior longitudinal ligament, PLL posterior longitudinal ligament, ISL interspinous ligament, LF ligamentum flavum, CL capsular ligament, AIL alar ligament, TL transverse ligament, ApL apical ligament, NL nuchal ligament

^a Values were taken from Table 1 in Zhang et al. [21], which were derived from the literature Ng and Teo [20]

Finite element model with two different anterior implants

All units representing the transverse ligament were removed from the cervical spine model to establish a finite element model of atlantoaxial instability. The anterior atlantoaxial transarticular locking plate system was drawn using Freeform Plus software and grid division was carried out and added to the finite element model of atlantoaxial instability according to the requirements of surgery. This established a finite element model to treat atlantoaxial instability with atlantoaxial transarticular locking plate system (Figs. 2, 3). The elastic modulus of the medical titanium was 110,000 MPa and Poisson's ratio was 0.30 [24]. Based on this established finite element model of atlantoaxial instability with the atlantoaxial transarticular locking plate system, all units of the titanium plate were then removed to establish a finite element model for the treatment of atlantoaxial instability using simple anterior transarticular screw fixation.

Boundary and loading conditions

Displacements of all the nodal points in the inferior margin of the axis on the X, Y, and Z axes were constrained to 0,

Table 2 Validation of the finite element model of the upper cervical spine (°)

	Panjabi et al. [23]		Normal model		Unstable model	
	C ₀ –C ₁	C ₁ –C ₂	C ₀ –C ₁	C ₁ –C ₂	C ₀ –C ₁	C ₁ –C ₂
Moment (Nm)	1.5		1.5		1.5	
Flexion	3.5 ± 0.6	11.5 ± 2.0	3.1	11.7	6.7	14.2
Extension	21.9 ± 1.9	10.9 ± 1.1	20.5	9.5	21.2	12.7
Lateral bending	5.6 ± 0.7	4.0 ± 0.8	5.1	4.1	6.1	6.2
Axial rotation	7.9 ± 0.6	38.3 ± 1.7	7.6	38.7	9.5	45.1

two loading points for flexion and extension were selected at the slope and the squamous part, two rotatory loading points were selected at both sides of the occipital condyle, and two lateral flexion loading points were selected at the occipital condyle. Forty Newtons of vertical downward pressure were imposed on the surface of the occipital condyle to simulate the weight of the head. Approximately 1.5 Nm torque was imposed on the model from various directions to produce flexion, extension, lateral bending and axial rotation. The models of the anterior atlantoaxial transarticular locking plate system and the simple anterior transarticular screw fixation were compared using an Ansys 10.0 postprocessor. The displacement of the lower margin of the C₂ vertebral body was set to 0, and the micromotion between the bone and the titanium plate and screws was ignored.

Stress diagrams and displacement diagrams were analyzed and compared. The quantitative method calculated the maximum displacement of the anterior transarticular screw (where maximum displacement includes both maximum flexion and extension angles), maximum paradigm of stress, and maximum paradigm stress of the screw–bone interface. Our model used a cylinder to represent the screw. The interface between screws and surrounding bone was defined as ‘close contact without movement’ in our Finite Element Analysis. The screws placed through the titanium

plate to the atlas and axis formed the simulated screw–plate and screw–bone interfaces. These interfaces were defined as close contacts, and their micro-movements were ignored.

Results

Displacement of two implants

The maximum displacement of the anterior transarticular screw is shown as it appeared during extension in both internal fixation devices (Fig. 4). The maximum displacements of the anterior transarticular screw at flexion, extension, lateral bending, and axial rotation states were 7.411×10^{-2} , 11.89×10^{-2} , 3.227×10^{-2} , and 2.501×10^{-2} mm, respectively. Maximum values of screw displacement were located at the screw tail. The anterior atlantoaxial transarticular locking plate system produced less displacement compared to the simple anterior atlantoaxial transarticular screw fixation by 24.6, 16.6, 16.4, and 36.8 % in flexion, extension, lateral bending and axial rotation, respectively (Fig. 4).

Von mises stress of two implants

During axial rotation (Fig. 5), stress concentration within the anterior atlantoaxial transarticular locking plate system was located at the point where the screw passes the atlantoaxial joint and screw holes, and it was similar in other motion states. As shown in Fig. 5b, the two axial vertebral screws bore relatively less stress at the junction of the screw and plate, and the screw body. There was no significant stress concentration.

From Fig. 6a, the maximum stresses on the transarticular screw were 5.213×10^1 , 1.029×10^1 , 1.164×10^1 , and 3.213×10^1 MPa during flexion, extension, lateral bending and axial rotation, respectively. The anterior atlantoaxial transarticular locking plate system produced less stress on the screw by 4.19, 2.83, 9.91 and 11.76 % in

Fig. 2 The finite element model of anterior atlantoaxial transarticular locking plate system **a** frontal view, **b** lateral view. Simulating Fig. 1, the columnar parts are two transatlantoaxial joint screws, and the wing-like structure is the locking titanium plate placed anterior to the axis vertebra

Fig. 3 **a** The finite element model treating atlantoaxial instability with atlantoaxial transarticular locking plate system. **a** Frontal view, **b** lateral view. The insertion point of the atlantoaxial transfacet screw is selected at the bony depression inferior to the anterior arch of the axis vertebra. The *upper* two screws are inserted into the lateral masses of the atlas in an inferomedial to superolateral direction to fix the atlas and axis and achieve an immediate stability between the atlas and the

axis. **b** The angle of the screws in anterior atlantoaxial transarticular fixation: **a** frontal view, **b** lateral view. The insertion point of the atlantoaxial transfacet screw is selected at the bony depression inferior to the anterior arch of the axis vertebra. The *upper* two screws are inserted into the lateral masses of the atlas in an inferomedial to superolateral direction to fix the atlas and axis and achieve an immediate stability between the atlas and the axis

Fig. 4 The peak displacement of two implants

flexion, extension, lateral bending and axial rotation, respectively (Fig. 6a), compared to the simple anterior atlantoaxial transarticular screw fixation.

Von mises stresses between the screws and vertebra of two implants

During flexion, extension, lateral bending, and axial rotation, the maximum Von mises stresses at the screw–bone interface of the transarticular screw were 1.761×10^1 , 3.568×10^1 , 5.610×10^1 , and 1.849×10^1 MPa, respectively. These stresses were less than those measured at the simple screw fixation group by 24.99, 15.62, 23.07, and 10.44 % in flexion, extension, lateral bending and axial rotation, respectively (Fig. 6b).

Discussion

As an effective internal fixation system using the antero-lateral retropharyngeal approach, the anterior transarticular screw has been shown to have a relatively poor resistance

Fig. 5 **a** The stress diagram of anterior atlantoaxial transarticular locking plate system under axial rotation. Scale bar reads 0.06 MPa (blue) through 16.1 MPa (red). Blue represents the minimum stress and red represents the maximum stress. **b** The stress distribution of

the C₂ screws of anterior atlantoaxial transarticular locking plate system under axial rotation movement. **a** Front view, **b** back view. Scale bar reads 0.06 MPa (blue) through 16.1 MPa (red). Blue represents the minimum stress and red represents the maximum stress

to flexion and extension. Kim et al. [7] have also shown that no significant difference was found between the posterior atlas lateral mass screw combined with axis pedicle screw, the posterior atlantoaxial transarticular screw, and the anterior atlantoaxial transarticular screw in three-dimensional motion. However, the screw displacement of the anterior transarticular screw is larger than the posterior atlas lateral mass screw combined with axis pedicle screw system, and this displacement is significant during flexion and extension. The anterior transarticular screw fixation not only can effectively solve such problems, but also provide a biomechanical stability equal to the Margel procedure for patients with atlantoaxial instability [25].

The main purpose of surgical intervention, in the case of atlantoaxial instability, is to obtain immediate stability and promote bony fusion of the atlantoaxial joint. Therefore, it is very important to evaluate the biomechanical properties of various atlantoaxial fixation methods. Our findings indicated that the anterior atlantoaxial transarticular locking plate system may provide a new option for internal fixation using the anterolateral retropharyngeal approach. The anterior atlantoaxial transarticular locking plate system

may also be used for certain cases which lack the posterior bony structures of the atlas and axis or in some cases requiring anterior spinal cord decompression. Moreover, it does not require extreme integrity of the atlantoaxial vertebral structure.

It was questioned whether rigid spinal internal fixation meant the emergence of stress concentration in internal fixation and a higher possibility of screw breakage [24]. At present, complications related to the anterior atlantoaxial transarticular screw fixation are rarely reported. Koller et al. [26] and Li et al. [27] and Reindl et al. [28] found no device-related complications in their case report. However, a 6.5–7.5 % incidence of screw breakage has been reported for posterior atlantoaxial transarticular screw fixation, which has similar biomechanical properties to the anterior transarticular screw [29, 30]. Results of the current study showed that the two screws of the anterior atlantoaxial transarticular locking plate system displayed smaller screw displacement and less stress at the screw–bone interface, which suggests that although the fixation strength increased in the anterior atlantoaxial transarticular locking plate system, the possibility of screw breakage did not increase.

Fig. 6 **a** The maximum Von mises stress of two implants, **b** the maximum Von mises stress between the screws and vertebra of two implants

The question of the possibility of screw breakage in the anterior atlantoaxial transarticular locking plate system, however, remains to be investigated in future studies. Different numerical studies on the effect of simulating rotational motions under load, or under displacement, or under hybrid control have shown that the nature of the boundary conditions strongly affect the mechanical evaluation of a simulated implant and the effect of this implant on the predicted biomechanics of a spine model [31, 32].

There is significant controversy surrounding the mechanical properties of the axis body screw. Kerschbaum et al. [33] reported that screw loosening occurred in two out of three initial cases of irreducible atlantoaxial dislocation treated with unicortical Harms screw–plate system. Yin et al. [34] applied their self-made TARP plate in the treatment of atlantoaxial instability and found that axis screw loosening occurred in approximately 10 % of cases. The anterior atlantoaxial transarticular locking plate

system produced less displacement compared to the simple anterior atlantoaxial transarticular screw fixation by 24.6, 16.6, 16.4, and 36.8 % in flexion, extension, lateral bending and axial rotation, respectively (Fig. 4); less stress on the screw by 4.19, 2.83, 9.91 and 11.76 % in flexion, extension, lateral bending and axial rotation, respectively (Fig. 6a); and less stress on the screw–bone interface by 24.99, 15.62, 23.07, and 10.44 % in flexion, extension, lateral bending and axial rotation, respectively (Fig. 6b). Although our study has shown that there was no significant stress on the axis body screw, further verification of its biomechanical properties is still warranted.

Our finite element analysis showed that the area of stress concentration within the anterior atlantoaxial transarticular locking plate system was mainly focused at the point where the screw passed the atlantoaxial joint and the screw holes (Fig. 2). To improve its mechanical properties, we applied the results of Lu et al. [35], who suggested that the screw insertion point of the transarticular screw should be the bony depression inferior to the anterior arch of the axis because its bone marker was obvious and the protruded bone (namely, anterior arch) in the depression near the joint could ensure sufficient bone mass in the axis. Therefore, the titanium plate was designed to be relatively thick to enhance the strength of the plate and make it suitable for screw fixation. To prevent breakage of the transarticular screw, bone grafting through the atlantoaxial joint surface is preferred to posterior bone grafting. However, implantation of cancellous bone particles is extremely difficult in this area because of the limited surgical exposure.

Finite element analysis is widely used for biomechanical analyses because it is capable of analyzing the inner stress and strain that are impossible or difficult to measure using ordinary experimental methods [36]. Comparison of our data with the screw displacement data obtained in cadaver tests performed by Panjabi et al. [23] suggests that the model established in the current study can effectively reflect upper cervical motion. To minimize the effect of model error on the accuracy of biomechanical evaluation, all units of the titanium plate were removed after establishing the anterior atlantoaxial transarticular locking plate fixation model to produce a pure anterior atlantoaxial transarticular fixation model for comparative studies.

Conclusion

In summary, we found that the three-dimensional screw displacement of the anterior atlantoaxial transarticular locking plate system may have an advantage with regards to stability. However, limited by our modeling approach to finite element analysis, our model defined the contact

between the bone and screw as a close contact (as no sliding contact model was used) and ignored the effect of micromotion on the bone–screw interface and bone absorption around the bone on the mechanical stability. The sliding contact model may better simulate internal fixation of the anterior atlantoaxial transarticular locking plate system and simplify the simple anterior atlantoaxial transarticular screw fixation method. This would further promote the accuracy of the finite element analysis method. Therefore, additional biomechanical experiments using the sliding contact model should be performed to verify the reliability of our new internal fixation system.

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Conflict of interest None of the authors has any potential conflict of interest.

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