Artificial atlanto-odontoid joint replacement through a transoral approach

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Received: 8 April 2008 / Revised: 31 October 2008 / Accepted: 10 November 2008 / Published online: 29 November 2008
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Abstract Resection of the odontoid process and anterior arch of the atlas results in atlantoaxial instability, which if left uncorrected may lead to severe neurological complications. Currently, such atlantoaxial instability is corrected by anterior and/or posterior C1–C2 fusion. However, this results in considerable loss of rotation function of the atlantoaxial complex. From the viewpoint of retaining the rotation function and providing stability, we designed an artificial atlanto-odontoid joint based on anatomical measurements of 50 pairs of dry atlantoaxial specimens by digital calipers and 10 fresh cadaveric specimens by microsurgical techniques. The metal-on-metal titanium alloy joint has an arc-shaped atlas component, and a hollow cylindrical bushing into which fits a rotation axle of an inverted v-shaped axis component and is implanted through a transoral approach. After the joint was implanted onto specimens with anterior decompression, biomechanical tests were performed to compare the stability parameters in the intact state, after decompression, after artificial joint replacement, and after fatigue test. Compared to the intact state, artificial joint replacement resulted in a significant decrease in the range of motion (ROM) and neutral zone (NZ) during flexion, extension, and lateral bending ($P < 0.001$); however, with regard to axial rotation, there was no significant difference in ROM ($P = 0.405$), a significant increase in NZ ($P = 0.008$), and a significant decrease in stiffness ($P = 0.003$). Compared to the decompressed state, artificial joint replacement resulted in a significantly decreased ROM ($P \leq 0.021$) and NZ ($P \leq 0.002$) and a significantly increased stiffness

($P < 0.001$) in all directions. Following artificial joint replacement, there was no significant difference in ROM ($P \geq 0.719$), NZ ($P \geq 0.580$), and stiffness ($P \geq 0.602$) in all directions before and after the fatigue test. The artificial joint showed no signs of wear and tear after the fatigue test. This artificial atlanto-odontoid joint may be useful in cases of odontoid resection due to malunion or nonunion of odontoid fracture, atraumatic odontoid fracture, irreducible atlas dislocation, posterior atlantoaxial subluxation, or congenital skull base abnormalities.

Keywords Artificial atlanto-odontoid joint · Biomechanics · Neutral zone · Range of motion · Transoral approach

Introduction

Damage to the odontoid process and transverse ligament of the C1–C2 complex may lead to atlantoaxial instability. Depending on its severity, patients may experience neck pain, limited neck mobility, sensory defects, or even gait abnormalities. Severe atlantoaxial instability may cause spinal cord compression resulting in paralysis or even death. Odontoid process resection may be necessary in case of malunion or nonunion of odontoid fracture, atraumatic odontoid fracture, irreducible atlas dislocation, posterior atlantoaxial subluxation, or congenital skull base abnormalities; the anterior decompression performed during surgical correction also results in atlantoaxial instability. This type of instability is currently corrected by anterior and/or posterior atlantoaxial fusion [11, 17, 19]. However, such fusion results in considerable loss of head and neck rotation and might adversely impact the quality of life of the patient.

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To address this issue, we have designed a novel, two-piece, artificial atlanto-odontoid joint that not only stabilizes the atlantoaxial complex, but also allows partial C1–C2 rotation and can be implanted onto the defect created by the anterior decompression. In the present study, we performed anatomical measurements on dry vertebral and fresh cadaveric specimens to determine the dimensions of the different parts of the artificial joint and the entry points and trajectories of fixation screws. After implanting the artificial joint, we performed biomechanical experiments to compare its stability with that in the intact state, decompressed state, and after fatigue test.

Materials and methods

Specimens

We studied 50 pairs of dry adult C1 and C2 vertebrae without skeletal abnormalities and 10 fresh cadaveric head and neck specimens extending from the top of the intact head to the C7 vertebra [male, mean age 39 years (range 23–57 years)]. General examination and X-ray images confirmed the absence of skeletal abnormalities. Dual-energy X-ray absorptiometry (QDR-2000; Hologic, Waltham, MA) of the specimens revealed a mean bone mineral density of 0.838 g/cm^2 (range, $0.692\text{--}0.963 \text{ g/cm}^2$); none of the specimens had osteoporosis. All specimens were sourced from our institutional anatomy department.

Specimen measurements

Measurements were performed using electronic vernier calipers (precision, 0.01 mm) and a protractor (precision, 0.1°). The dry specimens were measured to determine the midpoints of the atlas lateral masses (A and B in Fig. 1) and the midpoints of the axis vertebral body and lateral masses (C and D in Fig. 1) as entrance points of fixation screws for the atlas and axis vertebrae, respectively. Other measured variables of the atlas vertebra included the height and width of the anterior arch (including the anterior tubercle); the central length, width, and height of the lateral mass; and the lateral angulation of the lateral mass in relation to the sagittal plane (Fig. 2). For the axis vertebra, the other measured variables included the height and the maximum sagittal and transverse diameters of the odontoid process; the length, width, and height of the vertebral body; and the odontoid process retroversion angle between the coronal plane and a line passing through the centers of odontoid process and vertebral body (Fig. 3).

Using microsurgical techniques and a transoral approach, the head and neck specimens were dissected to reveal the layers of the posterior pharyngeal wall, the

Fig. 1 Measurements to determine the fixation screw entrance points for the atlas (A, B) and axis (C, D) vertebrae. The distance of the vertebral artery from the anterior midline of the atlas (a) and from the anterior midline of the axis (b) is indicated

Fig. 2 Atlas measurements. The lateral angulation (a) and length (b) of the lateral mass and the width of the anterior arch (c) of the atlas were measured

Fig. 3 Axis measurements. The retroversion angle (a) and sagittal diameter (b) of the odontoid process were measured

course of the vertebral artery, and the anatomical relationships of the atlantoaxial complex with the adjacent structures, carefully preserving the osteoligamentous tissues. The specimens were fixed in a surgical platform facing upwards, and the mouth was opened with retractors to obtain a clear view of the pharynx. Using microsurgical instruments, the individual layers of the posterior pharyngeal wall were incised to expose the atlas and axis vertebrae. Then, we measured the distance of the vertebral artery from the anterior midline of the atlas (a in Fig. 1) as well as from the midline of the axis vertebra (b in Fig. 1), and the thickness of the posterior pharyngeal wall overlying the anterior tubercle and lateral mass of the atlas and the vertebral body of the axis.

Design of the artificial joint

The artificial atlanto-odontoid joint was composed of a titanium alloy (Ti6Al4V; Northwest Nonferrous Metal Research Institute, China) and comprised atlas and axis components and cancellous bone screws. This metal-on-metal joint had highly polished areas of contact. The cancellous bone screws were 3.5 mm in diameter and 13–22 mm in length. The arc-shaped atlas component included a fixing plate and bushing (Fig. 4). The fixing plate was bilaterally symmetrical; had a thickness of 1.6 mm, height of 6.0 mm, and had screw holes at both ends. The hollow cylindrical bushing had a circular cross-section with an inner diameter of 4 mm, external diameter of 8.0 mm, and height of 6.0 mm. The axis component comprised an inverted v-shaped fixing plate, two cylindrical bedplates, and a cylindrical rotation axle (Fig. 5). The rotation axle had a diameter of 4.0 mm and fitted snugly into the atlas bushing. The height and diameter of bedplate 1 were 5.3 and 9.5 mm, respectively. Bedplate 2 had a diameter of 7.0 mm; its height varied according to the joint category (described later). Two processes projected outward at an angle of 45° from either sides of the midline of bedplate 2. The bedplates and processes of the axis component were integrated to prevent excessive rotation. The axis fixing plate had a thickness of 1.6 mm, and screw holes were present at the ends of its two arms. The axis fixing plate

Fig. 4 The atlas component of the artificial atlanto-odontoid joint. The fixing plate (A) and bushing (B) are indicated

Fig. 5 The axis component of the artificial atlanto-odontoid joint. The rotation axle (A), processes (B), bedplate 2 (C), bedplate 1 (D), and inverted v-shaped fixing plate (E) are indicated

along with bedplate 1 had a posterior angulation of 10° in relation to the coronal plane. The areas of contact between the bushing of the atlas component and the rotation axis and bedplate 2 of the axis component were highly polished. The artificial joint allowed unilateral (left or right) rotation of 45° via the atlas and axis vertebrae, but restricted the flexion, extension, and lateral bending. The artificial atlanto-odontoid joints were divided into three categories—large, medium, and small. The distance between the centers of the two screw holes of the atlas fixing plate, height of bedplate 2, height of rotation axis, and distance between the centers of the two screw holes on the axis fixing plate of the three categories of atlanto-odontoid joints were as follows: large, 35.7, 2.4, 8.1, and 23.2 mm; middle, 34.0, 2.3, 7.7, and 22.1 mm; and small, 32.3, 2.2, 7.3, and 21.0 mm, respectively.

Biomechanical model

Intact specimens

After the anatomical measurements, a segment of the vertebral column extending from the occipital condyle (C0) to C3 was excised from each of the ten cadaveric specimens and stored at −20°C in a sealed plastic bag until use. The specimens were naturally thawed at room temperature (20°C). Before the biomechanical tests, the muscles of the cervical specimens were resected, taking care to preserve the osteoligamentous structures.

Decompression specimens

The junction of the anterior arch and lateral mass of the atlas was carefully ground with a spherical drill bit (diameter, 1.0 mm) mounted on an electric grinder (Stryker System 2000; Stryker Corporation, MA). After grinding, the anterior arch of the atlas was completely removed, exposing the odontoid process. Grinding was carefully continued at the junction of the odontoid process with the axis vertebral body. The attachments of the apical, alar, and transverse ligaments to the odontoid process were carefully resected, and the odontoid process was completely removed, thus creating a biomechanical model of decompression.

Artificial atlanto-odontoid joint implantation

An appropriate category of artificial atlanto-odontoid joint was implanted onto the decompressed specimens. Close contact between the atlas and axis components was ensured by applying moderate vertical compression to these components; in addition, moderate extension (20°) of the atlanto-occipital joint was maintained. A Kirschner wire (diameter, 1.5 mm) was used to drill a hole at the predetermined entry points; the length of the Kirschner wire penetrating the bone determined the appropriate screw length. The atlas fixation screws were implanted parallel to the long axis of the lateral mass, with the screw trajectory directed 10° in the lateral direction. The axis fixation screws were implanted parallel to the cross-section of the vertebral body, with the screw trajectory directed 10° in the

medial direction (Fig. 6a, b). Figure 7 shows the artificial atlanto-odontoid joint in place after implantation over a decompressed specimen. Figures 8 and 9 show the trajectories of atlas and axis screws.

Biomechanical experiments

The range of motion (ROM), neutral zone (NZ), and stiffness of the cadaveric specimens were measured in the intact state, in the decompressed state, after artificial joint placement, and after a fatigue test. The ROM and NZ are important variables describing the three-dimensional motion of the spine. In the present study, the ROM was defined as rotation from the neutral position to a position of maximum load. The NZ was defined as rotation from the neutral position to the unloaded position at the beginning of the third load cycle [22, 28]. The C0 and C3 vertebrae were embedded in a special metal mould containing polymethyl methacrylate (PMMA) to maintain the specimen in the upright position at the center of the metallic mould with no rotation or lateral flexion. In this position, C0 had a posterior angulation of 20° , while C3 was parallel to the horizontal plane. The specimens were removed from the moulds after the PMMA had solidified. A total of six markers comprising 1-cm³ hollow plastic cubes were pinned onto the vertebra to simplify laser scanning and computer recognition. These markers were not in contact with each other and could be readily distinguished. The markers were scanned to describe the corresponding trajectory of the cervical segments [23]. A 3D laser-scanning measurement system (Real Scan USB Scanner 200;

Fig. 6 X-ray images (a, anteroposterior view; b, lateral view) of the artificial atlanto-odontoid joint implanted onto a decompressed specimen

Fig. 7 The artificial atlanto-odontoid joint implanted onto a decompressed specimen

Fig. 8 The atlas screws trajectories. The screws will typically be aimed 10° laterally in the axial plane

3-Dimensional Digital Corp., Danbury, CT) was used in the flexibility test to track segmental cervical motion under each testing condition. The precision of the system [i.e., standard deviation (SD) of the measurements] for rotation and translation was within 0.02° and 0.01 mm, respectively.

Spine testing was performed in a non-destructive manner. The C3-embedded blocks were fixed to the moving base of a 3D motion system, while the C0-embedded blocks were fixed to the top connector plate whose weight was counterbalanced through a suspension system to allow unrestricted motion of the experimental specimens in all directions. Some studies on the C1–C2 segments have suggested that a 1.5-Nm torque can produce the most appropriate physiological movement range, demonstrate pre- and post-experimental changes without damaging the specimen, and is less than the elastic limit of the cervical spine [2, 4, 5]. We conducted preliminary experiments to confirm these reports. With the connector loaded through the top of the experimental specimens, a multidirectional

Fig. 9 The axis screws trajectories. The trajectories of the screws are intended to be 10° medially in the axial plane

flexibility test was performed to determine the stability of the construct, applying pure moments of flexion, extension, right and left lateral bending, and right and left axial rotation in three equal steps to a maximum of 1.5 Nm. Each step used uniform loading, a translation loading rate of 5 mm/min, and a rotation loading rate of 15°/min. Each load step was maintained for 30 s to eliminate creep, and three load–unload cycles were used to precondition the specimens and minimize viscoelastic effects [8].

Stereophotographs and 3D laser-scanning measurements of the markers fixed to the specimens were taken only during the third load cycle. The ROM and NZ measurements of the atlantoaxial segments caused no fracture or ligament injury to the cervical specimens. The total measurement time for each specimen was less than 8 h. In the lateral bending and axial rotation experiments, the right and left lateral bending and axial rotation experimental steps were conducted in a random order. An MTS-858 biomaterials testing machine (MTS Systems, MN; precision, 0.01 Nm) was used for the stiffness and fatigue tests. The stiffness of the specimens was assessed during flexion/extension, lateral bending, and axial rotation (maximum torque, 1.0 Nm; speed, 15°/min). The fatigue test comprised 5,000 repetitions of axial rotation and 5,000 repetitions of flexion/extension performed in a random order (fatigue load, 1.0 Nm; frequency, 0.25 Hz). The ROM, NZ, and stiffness were measured before and after the fatigue test.

Statistical analysis

All results were analyzed using SPSS ver. 13.0 (Chicago, IL). The results were presented as ($\bar{x} \pm s$). The data of ROM, NZ, and stiffness were analyzed by *t* test. *P* values less than 0.05 were considered statistically significant.

Results

Tables 1 and 2 show the various measurements of the 50 dry atlantoaxial and 10 cadaveric head and neck specimens. Table 3 shows the atlantoaxial kinematics in the intact state, decompressed state, after artificial joint implantation, and after the fatigue test. Compared to the intact state, decompression resulted in a significantly increased ROM in all directions ($P \leq 0.002$). Furthermore, compared to the intact state, artificial joint replacement resulted in a significantly decreased ROM during flexion, extension, and left and right lateral bending ($P < 0.001$); however, there was no significant difference in the ROM during left and right axial rotation ($P = 0.405$). Compared to the decompressed state, artificial joint replacement resulted in a significantly decreased ROM in all directions ($P \leq 0.021$). Following artificial joint replacement, no significant difference in the ROM in all directions was observed before and after the fatigue test ($P \geq 0.719$).

Compared to the intact state, decompression resulted in a significantly increased NZ in all directions ($P < 0.001$) (Table 4). Furthermore, compared to the intact state,

Table 1 Measurements of 50 pairs of dry atlantoaxial specimens

Variable	Measured values	$\bar{x} \pm s$
Atlas		
Anterior arch height (mm)	8.3–14.2	11.1 ± 1.3
Anterior arch width (mm)	13.9–20.5	16.6 ± 1.7
Lateral mass length (mm)	13.7–22.1	17.8 ± 1.7
Lateral mass width (mm)	11.2–17.4	14.1 ± 1.5
Lateral mass height (mm)	10.3–17.1	13.5 ± 1.6
Lateral mass lateral angulation	3.5° – 17.5°	$11.3^\circ \pm 4.2^\circ$
Axis		
Odontoid process height (mm)	12.1–17.3	15.7 ± 1.1
Odontoid process sagittal diameter (mm)	8.9–13.1	10.9 ± 0.8
Odontoid process transverse diameter (mm)	7.9–11.9	9.3 ± 0.7
Vertebral body length (mm)	14.8–19.6	17.4 ± 1.7
Vertebral body width (mm)	15.9–23.1	18.2 ± 1.8
Vertebral body height (mm)	17.3–25.8	21.1 ± 1.8
Odontoid process retroversion angle	4.9° – 16.1°	$9.8^\circ \pm 2.1^\circ$

Table 2 Measurements of 10 cadaveric head and neck specimens

Variables	Measured values (mm)	$\bar{x} \pm s$ (mm)
Distance of vertebral artery from		
Atlas midline	20.4–29.7	25.6 ± 3.0
Axis midline	14.3–22.7	18.8 ± 2.5
Width exposed		
Atlas	37.6–43.2	39.7 ± 1.7
Axis	36.5–42.6	39.3 ± 2.1
Thickness of posterior pharyngeal wall overlying		
Anterior tubercle of atlas	2.8–4.1	3.5 ± 0.4
Lateral mass of atlas	5.0–7.4	6.6 ± 0.7
Vertebral body of axis	4.8–6.9	5.5 ± 0.6

artificial joint replacement resulted in a significant decrease in NZ during flexion/extension and right and left lateral bending ($P < 0.001$) and a significant increase in NZ during right and left axial rotation ($P = 0.008$). Compared to the decompressed state, artificial joint replacement resulted in a significantly decreased NZ in all directions ($P \leq 0.002$). Following artificial joint replacement, no significant difference in NZ in all directions was observed before and after the fatigue test ($P \geq 0.580$).

Compared to the intact state, decompression resulted in a significantly decreased stiffness in all directions ($P < 0.001$) (Table 5). Furthermore, compared to the intact state, artificial joint replacement resulted in a significantly increased stiffness during flexion, extension, and right and left lateral bending ($P \leq 0.001$) and a significantly decreased stiffness during right and left axial rotation ($P = 0.003$). Compared to the decompressed state, artificial joint replacement resulted in a significantly increased stiffness in all directions ($P < 0.001$). Following artificial joint replacement, no significant difference in stiffness in all directions was observed before and after the fatigue test ($P \geq 0.602$).

Discussion

To avoid the loss of rotation that follows odontoid process resection with anterior and/or posterior stabilization, we designed a new type of two-piece, metal-on-metal, artificial atlanto-odontoid joint which when implanted onto a decompressed specimen allowed the relocation of the axis of rotation to the artificial joint. Compared with the intact state, artificial joint replacement resulted in decreased ROM and NZ and increased stiffness during flexion, extension, and lateral bending. In the artificial joint, the rotation axle of the axis component fits snugly into the bushing of the atlas component because both parts have a

Table 3 Comparison of range of motion

Movement	Intact state	After decompression	After artificial joint replacement	After fatigue test
Flexion	11.6 ± 1.8	16.4 ± 2.2 ^a	7.1 ± 1.4 ^{a,b}	7.3 ± 1.5 ^{a,b}
Extension	11.3 ± 1.6	19.7 ± 2.4 ^a	7.6 ± 1.2 ^{a,b}	7.9 ± 1.4 ^{a,b}
Lateral bending	4.5 ± 1.1	9.2 ± 1.7 ^a	2.4 ± 0.8 ^{a,b}	2.5 ± 0.9 ^{a,b}
Axial rotation	43.1 ± 4.6	48.9 ± 5.1 ^a	44.3 ± 0.9 ^b	44.2 ± 1.0 ^b

Results are in degrees (°) and obtained with a 1.5-Nm load

^a $P < 0.05$ versus the intact state

^b $P < 0.05$ versus after decompression

Table 4 Comparison of neutral zone

Movement	Intact state	After decompression	After artificial joint replacement	After fatigue test
Flexion/Extension	4.2 ± 1.0	8.7 ± 1.9 ^a	2.3 ± 0.8 ^{a,b}	2.5 ± 0.9 ^{a,b}
Lateral bending	1.6 ± 0.4	3.1 ± 0.7 ^a	1.0 ± 0.3 ^{a,b}	1.1 ± 0.3 ^{a,b}
Axial rotation	28.3 ± 3.1	38.7 ± 4.2 ^a	33.1 ± 3.5 ^{a,b}	33.6 ± 3.7 ^{a,b}

Results are in degrees (°) and obtained with a 1.5-Nm load

^a $P < 0.05$ versus the intact state

^b $P < 0.05$ versus after decompression

Table 5 Comparison of stiffness

Movement	Intact state	After decompression	After artificial joint replacement	After fatigue test
Flexion	0.81 ± 0.17	0.41 ± 0.10 ^a	1.37 ± 0.26 ^{a,b}	1.32 ± 0.29 ^{a,b}
Extension	0.82 ± 0.19	0.38 ± 0.13 ^a	1.39 ± 0.28 ^{a,b}	1.33 ± 0.24 ^{a,b}
Lateral bending	1.82 ± 0.38	0.74 ± 0.17 ^a	2.88 ± 0.52 ^{a,b}	2.83 ± 0.48 ^{a,b}
Axial rotation	0.23 ± 0.05	0.08 ± 0.02 ^a	0.16 ± 0.03 ^{a,b}	0.15 ± 0.03 ^{a,b}

Results are in Newton-meter per degree (Nm/°) and obtained with a 1.0-Nm load

^a $P < 0.05$ versus the intact state

^b $P < 0.05$ versus after decompression

diameter of 4 mm. This limits flexion, extension, and lateral bending. While designing the joint, our primary aim was to facilitate C1–C2 rotation, which is the most important function of the C1–C2 complex. Moreover, even under physiological conditions, atlantoaxial flexion, extension, and lateral bending are actually quite limited. This is confirmed through the results of multidirectional flexibility experiments on the human cervical spine conducted by Panjabi et al. [16]. Indeed, head and neck flexion, extension, and lateral bending occur primarily as a result of the combined action of the atlanto-occipital joint and the cervical intervertebral joints after C2 (including the intervertebral discs and lateral mass joints). Thus, although our artificial joint would restrict flexion, extension, and lateral bending postoperatively, the actual overall impact on head and neck motion should be quite limited. We also found that compared to the intact state, artificial joint replacement and the fatigue test resulted in increased NZ of axial rotation and decreased stiffness. This could be due to

the loss of alar and transverse ligaments and other ligamentous structures during the decompression procedure.

The anatomical structure and ligaments of the normal atlantoaxial joint strictly limit excessive rotation in the physiological state. The normal range of C1–C2 rotation has been reported to be 50° on each side [20]. Cadaveric spine experiments have shown that complete bilateral atlantoaxial dislocation occurs beyond 64°–68° rotation [9, 13]. When artificial atlanto-odontoid joint replacement is performed after a decompression procedure through a transoral approach, there is loss of structures that limit excessive movement, such as the anterior longitudinal ligament, alar ligament, apical ligament of dens, and transverse ligament. Moreover, the lateral and posterior structures of the atlantoaxial joint place few restrictions on rotation. If the artificial joint merely had a simple rotation mechanism and no movement-restricting structures, excessive rotation could lead to fracture or dislocation of the lateral facet joints and injury to the spinal cord,

vertebral artery, and other important adjacent structures. Therefore, we included the two processes on bedplate 2; these processes limit unilateral rotation (left or right) to 45° , which is similar to the normal physiological range.

Both the fixing plate and bushing of the atlas component of the artificial atlanto-odontoid joint have a height of 6 mm; this is less than the height of the anterior arch of atlas (11.1 ± 1.3 mm), and would not influence flexion at the atlanto-occipital joint. The bushing has a diameter of 8.0 mm and bedplate 1 of the axis component has a diameter of 9.5 mm; these are less than the sagittal diameter of the odontoid process (10.9 ± 0.8 mm) and would avoid compression of the spinal cord. The design of bedplate 2 prevents collision between the atlas fixing plate and bedplate 1 during rotation. The axis fixing plate and bedplate 1 have a 10° posterior angulation; therefore, when the fixing plate is implanted onto the vertebral body of the axis, the position of the rotation axle will be similar to that of the odontoid process, which under physiological conditions has a retroversion angle of $9.8^\circ \pm 2.1^\circ$.

The distance between the vertebral artery and the midline of the atlas (25.6 ± 3.0 mm) was greater than that between the centers of the screw holes and the midline of the atlas fixing plate. Similarly, the distance between the vertebral artery and the midline of the axis (18.8 ± 2.5 mm) was greater than that between the centers of the screw holes and the midline of the axis fixing plate. However, it should be noted that during its course from the transverse foramen of the atlas to the foramen magnum, the vertebral artery passes through a groove present on the posterolateral margin of the atlas. Screws implanted into the atlas must not compromise this groove to avoid damage to the vertebral artery, and great care is essential intraoperatively to avoid damage to the bilateral vertebral arteries [1, 22].

The width (14.1 ± 1.5 mm) and height (13.5 ± 1.6 mm) of the atlas lateral mass and the width (18.2 ± 1.8 mm) and height (21.1 ± 1.8 mm) of the axis vertebral body are sufficient for unicortical fixation with a screw of 3.5-mm diameter. A previous study showed that bicortical fixation can significantly increase the stability of cervical specimens during flexion and lateral bending [12]. However, it could endanger the vertebral artery, spinal cord, and other important structures adjacent to the atlas lateral mass and axis vertebral body [15]. Unicortical fixation with cancellous bone screws is safer since the screws do not break through the posterior cortices of the atlas lateral mass or axis vertebral body.

The transoral approach, which was first described by Fang and Ong [6], allows excellent C1 and C2 exposure and complete spinal cord decompression [21]. We could expose the atlas over a width of 39.7 ± 1.7 mm, and the axis over a width of 39.3 ± 2.1 mm. These values are

greater than the widths of the atlas and axis components of the artificial joint, allowing adequate space for artificial joint placement. The thickness of the posterior pharyngeal wall was 3.5 ± 0.4 mm over the anterior tubercle of the atlas, 6.6 ± 0.7 mm over the lateral mass of the atlas, and 5.5 ± 0.6 mm over the vertebral body of the axis. Since the atlas and axis fixing plates are 1.6 mm in thickness, these are unlikely to induce posterior pharyngeal discomfort. Furthermore, the incidence of infections with the transoral approach has been now reported to be lower than that reported by Fang and Ong [6]. Although several other anterior approaches have been employed to access the upper cervical spine [3, 14, 18], the present transoral approach offers more intraoperative convenience, is less traumatic, and is associated with fewer postoperative complications.

After artificial joint replacement, testing for joint wear and fatigue is necessary to ensure its longevity [7]. We did not observe any notable difference in the ROM, NZ, and stiffness of the cervical specimens before and after the fatigue test. Moreover, none of the screws used for fixing the artificial joint had loosened. We had used the Light-Cutting Microscope (9J; Shanghai Optical Instrument Factory, China) measuring the finish on the surface of the rotation axle and the atlas bushing before and after the fatigue test, and had progressed contrast. The articular surfaces of the artificial joint retained their high-quality finish, with no signs of joint wear after the fatigue test. Nevertheless, in future experiments, we intend to design joints integrated with highly durable polymer components that are more resistant to wear and tear. To ensure the integrity of the specimens, we limited the testing of each specimen was completed within 8 h, the cycles of fatigue test were identified as 5,000 times during the entire tests. The results of our biomechanical experiments primarily reflect immediate stability. Furthermore, the degree of fatigue in our tests was considerably lower than that experienced under normal physiological conditions. It is as yet unknown whether this artificial atlanto-odontoid joint and its articular surfaces can maintain long-term stability with low joint wear. We will observe the long-term stability of the artificial atlanto-odontoid joint in animal experiment.

Hu et al. [10] have also developed an artificial atlanto-odontoid joint. In their joint replacement procedure, the anterior surface of the odontoid vertebral body is ground to create a bony channel into which the artificial odontoid component is fixed. However, that operative procedure is complicated and difficult to perform, the joint allows excessive rotation and provides less stability, there is greater local trauma and hemorrhage, and the results are not adequately supported by biomechanical tests. Our artificial atlanto-odontoid joint overcomes all these

shortcomings. However, we intend to conduct further experiments from the viewpoint of improving the durability and stabilization capacity of our artificial atlanto-odontoid joint for clinical application.

Conclusions

We have successfully performed anatomical measurements and designed a new type of artificial atlanto-odontoid joint for correcting atlantoaxial instability arising from C1–C2 anterior decompression procedures. The unique aspect of this joint is that it restores, to a great extent, the C1–C2 axial rotation that is lost during current stabilization procedures. Furthermore, the use of the transoral approach resolves the problems associated with other anterior approaches.

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