Three-Dimensional Measurement of Intervertebral Kinematics in Vitro Using Optical Motion Analysis

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Three-dimensional measurement of intervertebral kinematics *in vitro* using optical motion analysis

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Abstract: Measurement of the stiffness of spinal motion segments is widely used for evaluating the stability of spinal implant constructs. A three-dimensional motion analysis technique has been developed that allows accurate measurement of the relative movement of the vertebral bodies about a well-defined anatomical axis system. The position of marker clusters on each vertebra is tracked using digital infrared cameras (Qualisys AB, Gothenburg). Landmarks are identified using a marked pointer, and an anatomical coordinate system is defined for each vertebra. The transformation relating the upper and lower vertebrae is calculated, using the joint coordinate system approach of Grood and Suntay to find the rotations and translations in each anatomical plane.

The stiffness of vertebrectomy constructs was investigated using a Synex vertebral body replacement and an anterior rod with one or two screws in each vertebral body, with or without damage to the posterior longitudinal ligament (PLL). A moment of 2 N m was applied about each anatomical axis, and the range of motion about each axis was calculated.

The range of motion in flexion–extension and lateral bending was significantly greater with only one screw. When the PLL was cut, there was no significant increase in the range of motion.

Keywords: three-dimensional kinematics, intervertebral kinematics, spine, spinal construct stiffness, anterior and posterior spine instrumentation, vertebrectomy

1 INTRODUCTION

Spinal instrumentation is used extensively in the treatment of degenerative spinal pathologies. There are a number of clinical objectives, ranging from treatment to correct spinal deformities, to removal of damaged vertebrae or discs in order to reduce long-term back pain or spinal cord impingement, with a common post-operative aim of limiting the movement occurring between the affected levels of the spine, thus promoting fusion.

A variety of implant designs are currently available using either an anterior or a posterior approach. Posterior instrumentation generally consists of pedicle screws providing a rigid foundation for rods, which run between adjacent levels. Anterior instrumentation can take the form of cages replacing an excised intervertebral disc, of expanding cages replacing the vertebral body, or of screws and rods linking vertebral bodies.

Significant improvements have been made in measuring, understanding, and predicting spinal loading conditions as a result of the demands of the medical profession and their collaborations with spinal implant designers. The resulting implants can be used in various combinations with the aim of producing the most rigid spinal construct. With such a variety of implant designs now available it is important to know which are the most effective at limiting the movement of the spine. The most efficient order in which to insert the implants is also of interest to surgeons.

Nevertheless, there is as yet a lack of understanding as to whether implants create the correct conditions for spinal fusion in terms of construct stiffness and whether or how this stiffness is related

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to the fracture recovery process; i.e. by how much is intervertebral movement limited and to what extent should it be limited in order to promote fusion?

There are a great many biomechanical variables that need to be evaluated, and existing research has played a major role in improving the understanding of spinal loading using either apparatus developed to study spinal kinematics and kinetics or finite element models [**1–4**] to simulate the spinal construct and loading conditions. Some *in vivo* studies have investigated spinal kinematics [**5–9**], but it is difficult to measure intervertebral movement with sufficient accuracy to evaluate the effectiveness of implant constructs *in vivo*, and so most studies have been carried out on isolated motion segments *in vitro*. Most published studies have used straightforward static loading systems [**10–15**]; some researchers have developed more realistic loading systems [**16**, **17**], but this is difficult because of the complex and indeterminate muscle loads and ligamentous constraints.

The effectiveness of fusion and fracture fixation systems is typically evaluated by implanting the devices in animal or cadaveric motion segments, applying bending moments about the various anatomical axes, and measuring the resulting range of motion. More recently, axial follower loads have been used, in order to provide a closer approximation to normal physiological loading [**18**, **19**]. This simulates the stabilizing effect of compressive loading, although it could be argued that tests of the 'stability' of such constructs should simulate worst-case conditions where there is no compressive load holding the construct together.

A wide variety of different techniques have been used both to apply the loads to the constructs and to measure the resulting deflections. Loading systems range from simple weights [**10**, **11**] to sophisticated systems capable of applying complex multiaxial loading [**16**, **17**]. Methods of measuring the deflection of the construct include electrogoniometers [**13**], linear variable-differential transducers [**14**], and optical motion analysis [**15**].

A limitation of many of these systems is that they measure the angles of rotation of the vertebrae in two dimensions only. Coupled rotations are often not recorded, and the anatomical axis system is often assumed and not well defined, which may lead to significant errors, especially where there are relatively large movements. A further limitation is that the deflection is often measured through the loading apparatus, and any deficiency in the fixation of the vertebrae will lead to errors in the measured movements. Similarly, perfect fixation of (typically) the inferior vertebra to the apparatus is often assumed. It is not easy to fix the vertebral bodies rigidly, since the bone is soft and easily damaged under load, and so some errors are likely to occur here.

A three-dimensional motion analysis technique has been developed that allows accurate non-contact measurement of the relative movement of the vertebral bodies, independently of the loading apparatus, about a well-defined and repeatable anatomical axis system.

In this study, the stiffness of vertebrectomy constructs was investigated, using a vertebral body replacement supplemented with a single anterior rod, with either one or two screws in each vertebral body. The effect of damage to the posterior longitudinal ligament (PLL) was also assessed, since this is a common complication of vertebral body fractures.

2 MEASUREMENT TECHNIQUE AND ITS APPLICATION TO THE RELATIVE MOVEMENT OF VERTEBRAL BODIES

The measurement technique was based on a method developed by Holt *et al.* [**20**] to study tibiofemoral joint kinematics. Using digital infrared cameras (Qualisys AB, Gothenburg), the movement of marker clusters attached to each segment (using K-wires screwed into the vertebral bodies) is tracked with a sampling rate of 60 Hz. Within the laboratory global coordinate system, a coordinate system is defined by each marker cluster [marker coordinate system (MCS)], using the singular value decomposition approach of Soderkvist and Wedin [**21**]. Anatomical landmarks are identified on each segment using a marked pointer, and this allows an anatomical coordinate system (ACS) to be defined for each segment relative to the MCS. The transformation relating the upper and lower vertebrae can then be calculated, and the joint coordinate system approach of Grood and Suntay [**22**] is used to find the rotations and translations in the three anatomical planes.

The anatomical coordinate systems, and the definition of the axes of rotation, are shown in Fig. 1. The origin of each ACS was defined as follows: three anatomical landmarks are identified on the end plate of the vertebral body, the two most lateral points (X and X′) and the most anterior point (Y). The origin of the ACS (point O) is defined as the midpoint of the line connecting the most lateral points of the end plate, and this line also defines the x axis. The vector along OY is then calculated, and the z axis is defined by the cross-product of this vector and the unit vector in the x direction. The y axis is then defined

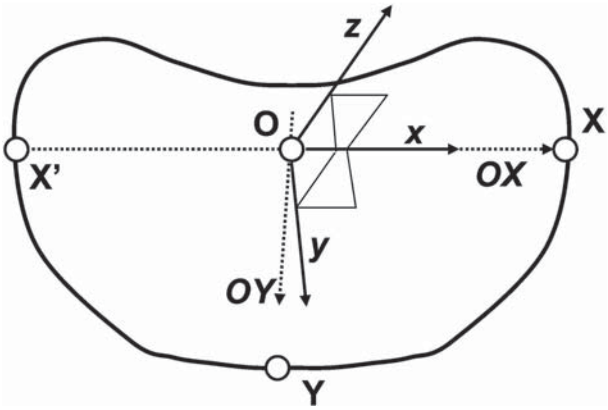

Fig. 1 Diagram showing the ACS, looking down on to the end plate of the vertebral body. Three anatomical landmarks are identified on the end plate of the vertebral body: the two most lateral points (X and X′) and the most anterior point (Y). The origin of the ACS (point O) is defined as the mid-point of the line XX′ connecting the most lateral points of the end plate. The x axis is defined by the unit vector $\boldsymbol{x}$ along OX. The vector along OY is then calculated, and $\boldsymbol{z}$ is defined by the cross-product of this vector and $\boldsymbol{x}$. $\boldsymbol{y}$ is then defined by the cross-product of the unit vectors $\boldsymbol{x}$ and $\boldsymbol{z}$

by the cross-product of the unit vectors in the x and z directions.

This method is used to define two axis systems, ACS1 in the superior end plate of the inferior vertebral body and ACS2 in the inferior end plate of the superior vertebral body. The rotation and translation of ACS2 relative to ACS1 are found using the joint coordinate system approach, where the flexion–extension axis is the x axis of ACS1, the axial rotation axis is the z axis of ACS2, and the lateral bending axis is a floating axis perpendicular to the other two. This approach has several significant advantages.

1. The axes correspond closely to conventional clinical terminology, so that the results are easily interpreted by clinicians.
2. The flexion–extension and axial rotation axes are defined by accurately measured anatomical landmarks, so that they are accurately positioned in the bones.
3. No singularities occur over the normal or abnormal physiological range of motion.
4. Joint displacements within this system are independent of the order in which the rotations and translations are specified, which is not commonly believed to be the case when using Euler angles.

For the purpose of this study, the position of the anatomical landmarks and the marker clusters were recorded for each motion segment in the unloaded neutral position. The positions of the marker clusters alone were then recorded in the neutral position before any loads were applied. Moments were then applied and the displacement of the marker clusters was measured in flexion, extension, left and right lateral bending, left and right axial rotation, and finally in the neutral position again. The range of motion in each anatomical plane was then calculated from the rotation and translation results.

Before testing the spinal constructs, the resolution of the measurement system was evaluated by attaching a marker to a micrometer stage and moving it in 0.005 mm steps. The positions of the marker were measured using the motion capture system.

Five cadaveric calf spine specimens (L3 to L5) were used. The calf spine has been shown by Wilke *et al.* [**13**] to provide a good model of the human lumbar spine and has the advantage that groups of spines are available from animals of similar age, weight, and sex that have had a similar diet and lifestyle. This allows much more reproducible results than using human spines, which are typically of poor and variable bone quality and condition.

In each specimen, the central vertebral body (L4) was removed to simulate a burst fracture and replaced with a Synex expanding cage (Synthes, Welwyn Garden City, UK). This was augmented with a Ventrofix (Synthes, Welwyn Garden City, Herts, UK) anterior rod system, fixed with two screws in each vertebral body (Fig. 2). The marker clusters were attached to each segment using K-wires screwed into the vertebral bodies. After the range of motion of each construct had been measured, the superior and inferior screws (Fig. 2) were removed, leaving one screw in each vertebral body, to simulate single screw fixation, and the measurements were repeated. Finally, the PLL was cut to simulate burst fracture injury or intraoperative damage, and the measurements were repeated again. Since all other aspects of each construct remained unchanged, this method allowed the effects of each change to be evaluated in isolation.

Mounting plates were attached to the motion segment using bone cement and two 6.5 mm cancellous AO screws in each vertebral body. To load the constructs, a simple system of weights was used, attached by cords to six quadrants fixed to the upper mounting plate, as shown in Figs 3 and 4. This maintained a constant-moment arm as the construct rotated. A moment of 2 N m was used, which is sufficient to measure the stiffness of the construct accurately in each direction without causing damage or excessive viscoelastic deformation that might

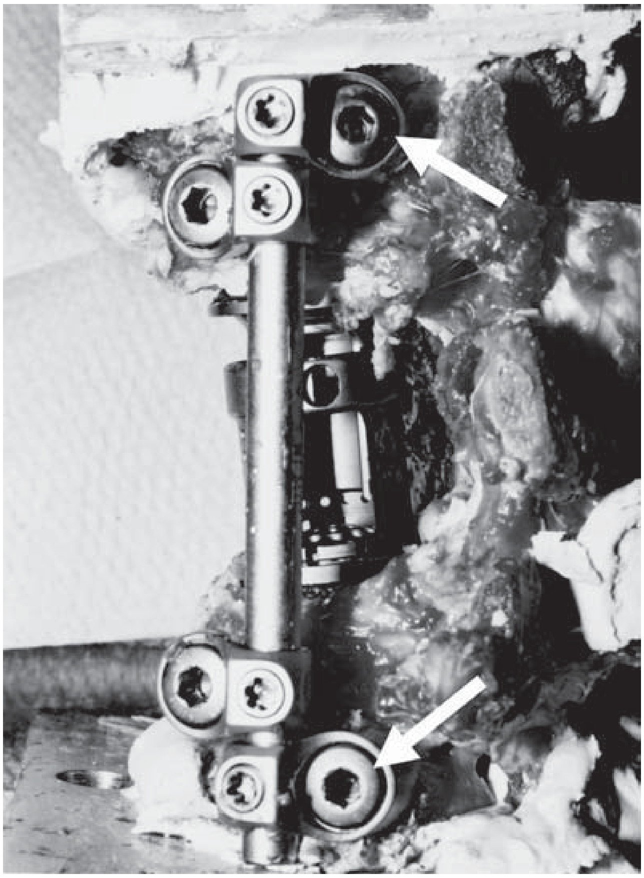

Fig. 2 Lateral view of a typical specimen showing the location and orientation of the cage, screws, and rods. The central vertebral body (L4) was removed to simulate a burst fracture and replaced with a Synex expanding cage (Synthes, Welwyn Garden City, UK). This was augmented with a Ventrofix (Synthes, Welwyn Garden City, Herts, UK) anterior rod system, fixed with two screws in each vertebral body. The white arrows indicate the screws that were subsequently removed to simulate single screw fixation

affect the results of subsequent measurements. While a moment is applied about one axis, the other cords are slack so that no moments are applied about the other two axes. Moments due to the movement of the centre of gravity of the quadrant assembly may be introduced, but, since the assembly is relatively light and close to the centre of rotation, the moment that this produces is negligible. The loading assembly was not aligned with the centre of rotation, but, since pure moments are applied and the upper vertebra is free to move in all six degrees of freedom, the effect is to produce a pure moment about the instantaneous centre of rotation.

The significance of the results was evaluated using paired two-tailed t tests to compare the constructs with one and two screws and those with the PLL

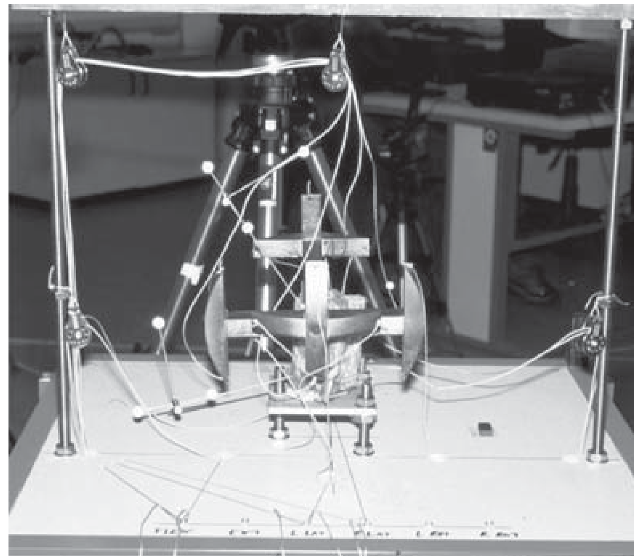

Fig. 3 Photograph showing the test set-up. To load the constructs, a simple system of weights was used, attached by cords to six quadrants to maintain a constant moment arm as the construct rotates. Ball bearing blocks (Harken, USA) minimize friction at the turning points. Using digital infrared cameras (Qualisys AB, Gothenburg), the movement of marker clusters attached to each segment is tracked with a resolution of 25 μm and 60 Hz sampling rate

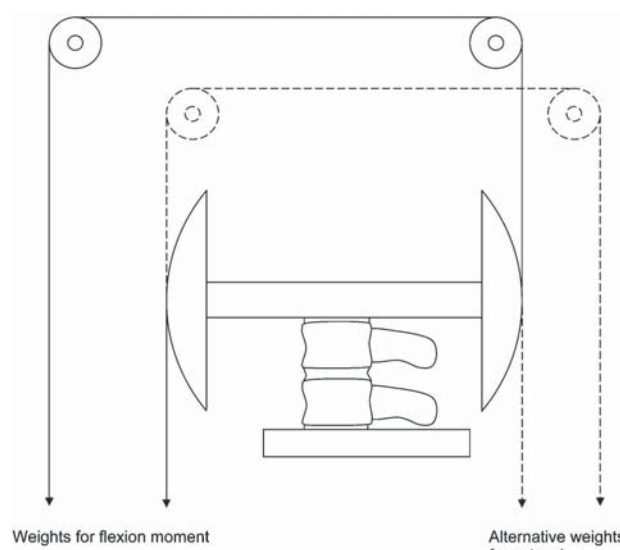

Fig. 4 Schematic diagram showing the operation of the loading apparatus for applying flexion–extension moments; moments about the other two axes were applied in a similar way, as seen in Fig. 3. The diagram shows a pair of weights for each loading direction, attached by cords to quadrants. The cords wrapped around the quadrants to maintain a constant-moment arm as the specimen rotated

intact and cut. Since the same spine segments were used, with the second screws being removed and the PLL cut *in situ*, it was possible to use a more powerful paired comparison that reduces the effects of the variation between spine specimens.

3 RESULTS

The minimum movement of a single marker that could be detected by the motion capture system was found to be 25 μm.

The results of the measurements are summarized in Fig. 5, which shows the mean range of motion about each axis together with the standard deviations. The range of motion with two screws was significantly smaller than with a single screw in flexion–extension ($p = 0.037$) and lateral bending ($p = 0.045$), but not in axial rotation ($p = 0.076$). Cutting the PLL made no significant difference to the construct stiffness about any of the three axes ($p > 0.05$).

Figure 6 shows the rotation about all three axes of a typical motion segment when loaded in axial rotation. It is evident that removing the second screws allowed substantial coupled rotation, especially about the flexion–extension axis.

4 DISCUSSION

The accuracy of the measurement system is dependent on the resolution of the motion capture system and the identification of the anatomical landmarks. The resolution of the camera system was 25 μm, which is better than most optical tracking systems used for gait analysis. This is due to two factors:

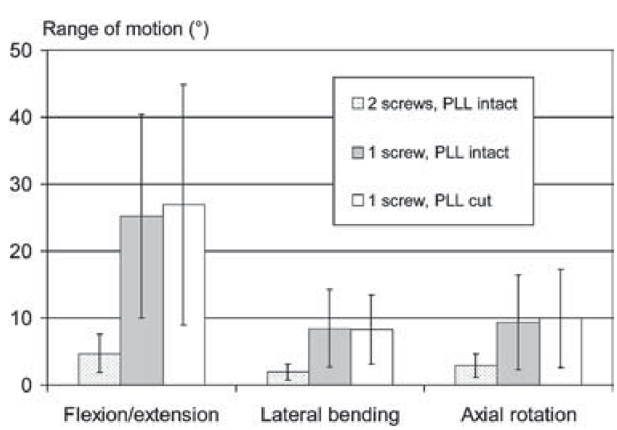

Fig. 5 Mean range of motion for each construct about each axis (the error bars show one standard deviation for each group)

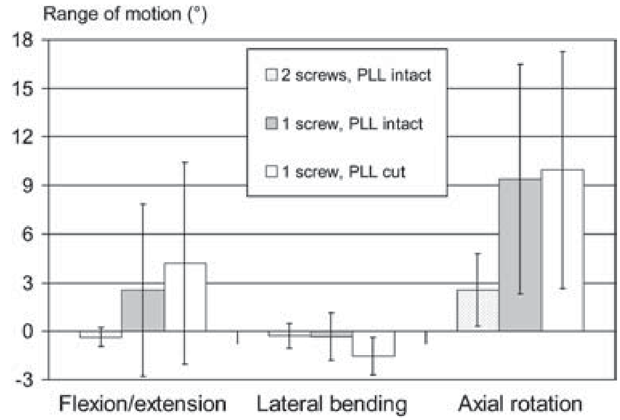

Fig. 6 Range of motion about each axis when loaded in axial rotation (the error bars show one standard deviation). When the second screw was removed, greater flexion–extension movement occurred, but not always in the same direction. This coupled flexion–extension movement allowed greater axial rotation to occur when an axial rotation moment was applied

firstly, the cameras were positioned much closer to the specimen than is usual for gait analysis, giving improved accuracy over a smaller field of view; secondly, the Qualisys ProReflex cameras use a proprietary subpixel interpolation algorithm that increases the resolution.

The resolution with which the angles of rotation are measured depends on the size of the marker clusters and the resolution of the motion capture system. For a marker cluster 70 mm across, the angular resolution is better than 0.05°.

The definition of the anatomical coordinate system is important. If an anatomical landmark is incorrectly identified, this will cause an angular movement of the anatomical axes and will affect the way that a movement appears as a set of rotations. Therefore, for each individual specimen, the positions of the landmarks were identified and marked as precisely as possible by cutting notches in which the pointer could be repeatably located, before assembling the construct. There was an inherent variation in the alignment of the axes between the specimens owing to differences in bone morphology, but the method of definition of the axes ensured that they were orthogonal and anatomically correct. It is important to note that no errors were introduced by lack of repeatability in defining the landmarks since the anatomical calibration was carried out only once for each specimen, and it was not necessary to repeat it when the screws were removed and the PLL was cut.

There appear to be no other studies in the literature that have compared the effects of using single or double anterior screws on vertebrectomy construct stiffness. In the present study, using two screws in each vertebral body produced a significantly stiffer construct than using a single screw. The screws ran laterally through the vertebral bodies, and, since a single screw can rotate in the bone, this allowed some flexion–extension movement, which was not possible where two screws were used. With a single screw, there was also significantly more movement in lateral bending, and a larger range of motion in axial rotation, although this difference was not statistically significant. This movement is attributed to coupled rotations of the vertebral bodies so that, when they were loaded about the lateral bending or axial rotation axes, some rotation about the other axes also occurred. Thus, the use of a single screw allowed more movement even when the construct was not loaded about the screw axis. This effect is illustrated in Fig. 6, which shows the rotations about all three axes when loaded in axial rotation. Removing the second screw allowed much more flexion–extension movement, resulting in a much increased range of axial rotation.

The mean flexion under axial rotation loading was not significantly different with one screw instead of two in each vertebral body (Student's t test, $p = 0.47$), but closer examination of the results revealed that this was because some constructs extended while others flexed, so that the mean was not greatly different. Fisher's F ratio was therefore used to test the significance of the difference in variance between the two groups, and this was found to be highly significant ($p = 0.000\,47$). The difference in variance was also significant between the group with two screws and the group with one screw and no PLL ($p = 0.000\,245$), but cutting the PLL made no significant difference to the group with one screw ($p = 0.38$). This confirms that a greater range of flexion–extension movement occurred when the second screw was removed, even though the construct was not loaded about the screw axis.

The different flexion–extension movements that were observed may also explain why the range of axial rotations was not significantly different ($p = 0.076$). Depending on the precise location of the screws and the anatomy of the motion segment, either flexion or extension occurred to a greater or lesser extent. As a result, the effect of removing one screw on the range of motion in axial rotation varied in different specimens. Although the mean range of motion was much greater with one screw rather than with two, this greater variability meant that the difference was not significant. However, the variances were significantly different (Fisher's F ratio, $p = 0.021$), because in some specimens the position of the single screw allowed much more axial rotation.

No attempt was made to distinguish between flexion and extension movements, or to measure asymmetry in lateral bending or axial rotation, because there was no clearly defined neutral position. The motion segments did not return to a consistent repeatable position when unloaded. This was confirmed by repeating the unloaded measurement before and after loading for each condition.

The large range of motion about each axis that was observed with only a single screw would clearly not be conducive to successful fusion of the construct and could represent a potentially dangerous instability if it occurred *in vivo*. The presence of an axial compressive load would help to stabilize the construct, and so the range of motion *in vivo* could possibly be expected to be smaller than was measured in the current experimental set-up. However, these results suggest that two screws should be used if possible.

The PLL was cut to simulate the possible effects of a burst fracture or an intraoperative injury, but it was found that this did not significantly affect the range of motion in any direction. Although the PLL is very important in stabilizing some fusion constructs, in this case the screw-and-rod system restricts extension and distraction and so the presence or absence of the PLL was less important. This is reassuring, since the PLL is commonly damaged in severe burst fractures.

The use of true three-dimensional measurement of the relative motion of the vertebral bodies was found to be valuable in properly measuring and understanding the complex coupled rotations that occurred in these constructs. Similar studies in the literature have typically only measured motion about one axis at a time, and, since the motion segments always exhibited coupled rotations about more than one axis, this would have been a significant limitation in the present study. Although a simpler measurement technique could have identified the benefit of using two screws, the present approach gave valuable additional insights into why a single screw allows movement in all directions and not just about the screw axis.

A further consideration is that, since coupled rotations occur, it is important to define the measurement axes accurately if the angles are to be measured correctly. The identification of reproducible anatomical axes in clinically relevant positions is important if such errors are to be avoided. The optical motion analysis system used in this study measured the relative movement of the bones independently of the loading mechanism, and this is also important since it is difficult to attach the loading mechanism rigidly to the soft vertebral bone.

5 CONCLUSIONS

The use of a single screw in each vertebral body allowed significantly more movement and produced a less stable construct than using two screws. A single screw can rotate in the bone; this allows a greater range of motion about all three axes owing to the complex coupled rotations that occur between the vertebrae.

The use of an optical measurement system that allows true three-dimensional measurement of the relative position of the vertebral bodies, about well-defined axes, and independently of the loading system, is valuable in this type of study.

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APPENDIX

Notation

ACS	anatomical coordinate system
MCS	marker cluster coordinate system