Thoracolumbar burst fracture is a common spine injury representing ~20% of thoracolumbar injuries. Favorable results were reported after non-operative management without reduction; however, surgical treatment is advocated in most cases to provide stability, prevent neurological deterioration, and provide pain relief. Burst fractures usually result from flexion-axial forces. However, such injuries require large forces that also disrupt soft tissues that stabilize the spine. Such injuries are usually result from massive compressive force that acts axially with the body excessively bending forward quickly due to rapid deceleration. This leads to the inherent instability, which can then lead to spinal cord compression and transaction.

The choice of surgery may vary according to surgeon’s preference and local product availability. With this in mind, the pedicle posterior fixation was suggested as the best option since it counteracts flexion forces. Posterior fixation involves the use of pedicle screws and longitudinal rigid rods that connect the screws. These implants provide the required stiffness to stabilize the damaged spine segment. The use of short segment posterior instrumentation was reported to be the ideal treatment as it preserves motion in the adjacent spine while preserving alignment.

However for comminuted fractures of the vertebra, this fixation is not recommended due to the lack of support and stability. In such cases, the use of long segment instrumentation, that is at least two vertebrae above and below the fracture, may be necessary to provide better stability. Although rod and screw systems enhance sagittal plane stiffness significantly in an injured spine, they may prove inadequate under lateral bending and axial rotation loading. The concept of using cross-links was established to overcome this issue. Their use was originally designed to provide greater correction and stability in scoliosis, particularly in the coronal plane. Subsequent studies found that they also improve stability during axial loading. Since then, cross-links between rods has been frequently used for posterior fixation procedures. Cross-linking might also improve construct stability and prevent rod migration and excessive lateral movements. Since fewer screws are required, cost is also reduced. Thus, the question remains: which cross-link configuration provides the best stability for thoracolumbar burst fracture?

Several studies showed promising results using cross-link configurations with posterior instrumentation. Dick et al. demonstrated that the use of horizontal links in segmental pedicle screws led to an increase in multiplanar stiffness. Conversely, Valdevit et al. demonstrated that the use of diagonal cross-link provided the most stable construct when rods were supplemented with or without transverse links. Wood et al. tested various scoliosis constructs to determine torsional stability of pedicle screws and cross-links, concluding that horizontal connectors increased stability.
Lim et al.\textsuperscript{11} found that cross-linking improved construct stiffness in all planes. Although a horizontal link provided a stiffer construct in the upright position, the oblique construct was stiffer when the spine was in flexion or extension.

Nonetheless, direct comparisons in stiffness among these constructs have not been demonstrated. Moreover, the assessment of multiple levels vs. single level fusions in burst fractures using the enhanced stability provided by cross-links has not been elucidated. We explored the possibility of using an X-shaped cross-link configuration for improved stability in surgical treatment of thoracolumbar burst fracture.

**MATERIALS AND METHODS**

**Geometry Reconstruction**

A 3D finite element (FE) model of the human thoracolumbar spine was reconstructed from 2D CT images of a healthy 40 year-old using image processing software (MIMICS 10.01, Materialise, Belgium). The CT data contained 1mm slices with pixel size of 0.742 mm. Element size was 0.5 mm for the entire model. A segmentation process was conducted on each slice to extract bony surface area (Fig. 1a). The final FE model consisted of T11-L3 vertebrae (cancellous bone, cortical shell and posterior bony elements), intervertebral disc (annulus fibrosus and nucleus pulposus), facet joints, major spinal ligaments and the posterior stabilization instruments. Material properties are listed in Table 1.

Facet joints and discs were not completely visible in CT images. Therefore, to model the cartilage, the extracted profiles of articulating surfaces were extruded with a thickness size that was obtained by determining the maximum distance between two bones. A Boolean operation was then performed for overlapping subtraction to achieve surface contact.\textsuperscript{18,19}

Due to the low resolution of the CT datasets, an automatic smoothing procedure was utilized to remove sharp edges in the model without changing geometry. After creating the segments of a fixed thoracolumbar spine, adjacent components were merged to create perfect surface contact. FE meshing was done with optimum mesh size for convergence; we applied a 0.1 mm mesh size for facet joints and discs and 0.5 mm for vertebral body. Seven ligaments: anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), intertransvers ligament (ITL), capsular ligament (CL), ligamentum flavum (LF), interspinous ligament (ISL), and supraspinous ligament (SSL) were simulated as linear tension spring elements\textsuperscript{20,21} (Fig. 1b).

**Modeling Thoracolumbar Spine Burst Fracture**

To simulate the burst fracture, a complete corpectomy was applied to the model by removing all vertebral body and adjacent intervertebral disc at the L1 level. The choice of an L1 fracture was based on previous evidence describing this region as a common fracture site.\textsuperscript{22} Furthermore due to inherent biomechanical properties at T12 and L1, mechanical forces result in the highest stress concentration in this region.\textsuperscript{23}

To mimic post-surgical conditions, ALL, PLL, and ISL were subtracted from the fixed model, similar to what is performed to implant a spinal cage. Then, a cylindrical cage and posterior instrumentation were constructed using 3D modeling software (Dassault Systèmes SolidWorks Corp, Boston, MA). The length and outer diameter of the pedicle screws were 45 and 4.5 mm, respectively, and the outer diameter of rods were 6 mm. The cage, screws, and rods with cross-links were then positioned in the corpectomy site with final approval provided by a spine surgeon (Fig. 1k). To model the polyaxial screw, the connection between the screw and its head was modeled as deformable—deformable, while the rod was rigidly connected to the head, allowing the head and rod to move relative to the screws. Finally, eight posterior fixation configurations were modeled: a long construct without cross-link (NL), a long construct with horizontal link (HL), a long construct with diagonal link (DL), a long construct with an X-shaped link (XL), a short construct with no link (NL), a short construct with horizontal link (HL), a short construct with diagonal link (DL), and a short construct with an X-shaped shape link (XL) (Fig. 1c–j). Long construct is defined as pedicle screw fixation involving T11-L3 while short construct is defined as fixation involving only T12–L2. In both, two rods were used to connect the head of the screws that followed the curvature of the normal thoracolumbar spine. Pedicle screws were assumed to be rigidly bonded to the bone. This technique assumes that micromotion between the screws and bone is negligible. The rotational displacement between T11 and L3 vertebrae and the Von Mises stress of the adjacent vertebra were analyzed.

**Analysis**

The FE models included from 155,157 to 202,180 nodes and from 653,805 to 833,835 elements. The model was loaded with compressive force of 200 N axially,\textsuperscript{20,34,35} and a 7.5 Nm moment on the superior surface of T11 to simulate the four physiological motions/functions of the spine.\textsuperscript{36,37} The inferior side of the L3 vertebral body was rigidly fixed.

The biomechanical effects of HL and DL were analyzed and compared to the NL model. A radical design of an X-shaped crosslink (XL) was also evaluated. Spine range of motion (ROM) (i.e., spine deformation spine after loading)\textsuperscript{29} with 7.5 Nm and Von Mises stress on the vertebrae were evaluated. A static FE analysis was performed with Marc (MSC Software, Santa Ana, CA). Sample size and significance were extrapolated from data obtained in our study and utilizing standard deviations from the literature for similar experimental models.\textsuperscript{12} All tests of significance were done at the p < 0.05 level using the Statistica software, Version 6.0 (StatSoft, Tulsa, OK).

**RESULTS**

Eight types of posterior pedicular screw fixation under four load conditions resulting in 32 FE models were evaluated.

**Validation of the Intact Model**

ROM of the model was compared with results of experimental studies in similar loading conditions\textsuperscript{39,40} (Fig. 2). In flexion, extension, lateral bending, and axial rotation, the model data corresponded to experimental data. The calculated ROM for T12-L2 in flexion, extension, bending and rotation were calculated as 7.7, 4.6, 12.5, and 6.5, respectively.

**Axial Displacement of the Entire Model**

In flexion (Fig. 3), long constructs using cross-links were significantly (p < 0.05) stiffer compared to the construct with no cross-links. This however, was not
observed among short constructs. The intact stiffness during flexion was 3.7 Nm/° and peaked at 15.12 Nm/° and 80.74 Nm/° for two-level and four-level constructs (with no cross-links), respectively. Adding additional cross-links increased stiffness in all constructs. The HL and DL cross-links used in long constructs were stiffer (88.14 and 89.76 Nm/°, respectively). However, the use of an XL link provided even higher stiffness.

Figure 1. Development of thoracolumbar FE model. (a) Segmentation to define separate vertebral bodies from CT. (b) Complete model of the intact spine T11-L3 includes major ligaments, ALL, PLL, ITL and CL, LF, ISL, SSL from front and lateral views. (c) Short construction N. (d) Short construction with a central HL. (e) Short construction with central DL. (f) Short construction with central XL. (g) Long construction NL. (h) Long construction with a central HL. (i) Long construction with central DL. (j) Long construction with a central XL. (k) Screw positioning in vertebral body.
The stiffness of a short construct with HL, DL, and XL was 15.93, 15.95, and 15.99 Nm/˚, respectively.

In extension, among the short constructs, that is the HL (9.45 Nm/˚) or DL (9.46 Nm/˚), and NL (9.43 Nm/˚), no significant differences were observed. However, the XL short construct significantly improved stiffness (from 9.43 to 10.54 Nm/˚).

In the long constructs, cross-linking provided significantly higher stiffness (Fig. 3). Increases from 49.24 to 54.52, 56, and 58.66 Nm/˚ were observed when using HL, DL, and XL cross-links, respectively.

In lateral bending (Fig. 3), the stiffness of the intact spine was 2.8 Nm/˚. A significant increase was observed when XL links were applied to long constructs. In contrast to flexion and extension, HL cross-link constructs were stiffer than DL constructs during bending. Using cross-links in short constructs also increased bending stiffness, although not significantly.

In axial rotation testing (Fig. 3), the stiffnesses of the intact spine, and NL constructs for short and long assembly were 2.53, 13.20, and 25.50 Nm/˚, respectively. Adding cross-links to long constructs increased stiffness significantly. The stiffnesses of the HL (31.81 Nm/˚), DL (29 Nm/˚), and XL (35.76 Nm/˚) with long constructs were significantly higher than that of the NL constructs (25.5 Nm/˚), while no differences were observed in short constructs.

Von Mises Stress on the Vertebral Body

The Von Mises stress on the T12 vertebra was greatest during axial rotation in the bone-screw interface. Stress in axial rotation with long construction without cross-link (NL) was 20 MPa and reduced to 15.3, 15, and 10.5 MPa with the use of HL, DL, and XL cross-links, respectively. XL reduced T12 stress by 30.6% against the DL cross-link during axial rotation. However, stresses for short constructs were 23.4, 17.4, 17.5, and 17.2 MPa with NL, HL, DL, and XL cross-links, respectively. Although cross-linking reduced the stress on T12, no significant differences were observed among configurations (Fig. 4a). For L2, the maximum values of Von Mises stress was observed in lateral bending (Fig. 4b). In long constructs without links (NL), L2 was subjected to 36 MPa. This value reduced to 18, 16, and 12 MPa when using HL, DL, and XL,

![Graph](image)

**Figure 2.** Comparison between the current FE model and experimental studies.
respectively. The HL, DL, and XL reduced bending stresses on L2 by 50%, 55.5%, and 66.7% as compared to NL constructs. In short constructs using cross-links, stress on L2 was reduced but not significantly. The stresses for NL, HL, DL, XL were 22, 17.5, 17.1, 17 MPa, respectively.

Von Mises Stress on Pedicle Screw Neck

Screws in long constructions experienced higher stresses than those in short constructs (Fig. 5). Cross-linking decreased stresses on the screw neck for short and long structures. During lateral bending, stresses were reduced for long NL from 133 to 120, 118, and 100 MPa in HL, DL, and XL constructs, respectively. During axial rotation in the long constructs the stresses reached 131, 113, 107, and 98 MPa in the NL, HL, DL, and XL constructs, respectively. In long fixation, stress was lowest in the XL configurations followed by DL and HL for all loading conditions. In axial rotation and lateral bending with XL configurations, stresses were reduced 25% and 26%, respectively. A reduction in stress was observed in short constructs with cross-links, but the values were not significant.

DISCUSSION

We analyzed the biomechanical changes and stability of posterior spinal fixation with the aid of cross-links in burst fractures involving the L1 vertebral body. The FE model was unable to precisely model features of the spine such as the vertebral cortex (thickness from 150 to 350 μm), and the inter-trabecular distance...
A uniform isotropic material was assumed for cancellous and cortical bone; however, both have been previously modeled as anisotropic or with rigid body elements. Of the eight configurations examined, the long construct using our proposed X cross-link provided the highest stability while providing the lowest stresses at the adjacent vertebral body and implants under all the tested loading conditions. For the short construct, cross-linking provided no benefit in stability and stress distribution. The reason for this remains unexplained in our study; however, several hypotheses can explain this finding. First, in the short construct, the use of only four pedicle screws was insufficient to provide a stable construct. This resulted in inherent instability since the strength of the construct now depended on the partially affected spine body. Second, in the short construct, the screws were inserted at vertebrae that were unsupported at L1. In the long construct, at least four pedicle screws were supported by vertebrae that are not adjacent to the affected spine, thus having better stability and support. This condition may have created a more stable cross-link construct since the ends of the cross-link bar were supported by strong screw attachments.

In our study, construct stability was reflected by increased stiffness. Therefore, the term “construct stability” is used interchangeably with “construct stiffness,” the latter being measured using the technique described in our study. In a clinical study, the group with rigid pedicle screw instrumentation had an increased fusion rate. While the lumbar spine allows flexion-extension, in the present scenario, the involvement of L1 had a major influence on the function of T12 and, in the long construct model, T11. Axial rotation is also an important parameter to be considered. In a study involving FE and experimental analyses, no differences were observed when cross-links or segmental pedicle screws in two-level constructs were subjected to axial rotation. However, the increase in the number of horizontal cross-links results in increased stability. Other authors also showed that adding cross-link devices only increased construct stability in rotation but had no significant impact in flexion, extension, and lateral bending. Our data demonstrate that diagonal and horizontal cross-links increase stability significantly, but our proposed X cross-link provided the best stability. The reason for improved rotational stability over other motions using cross-links remains elusive, but may be due to the angulations by which the rods were placed in relation to the vertical alignment of the spine.

In most clinical reports, the pedicle screws were regarded as the weakest link of the posterior construct. Our study demonstrates that the forces from physiological loading are concentrated at the screw neck and the contact between the screw and the lamina as reported in the literature. Implant
failures usually result from screw loosening or screw neck breakage. Chen et al.\textsuperscript{52} demonstrated that even after fusion, the pedicle screw continues to take a large portion of the load and therefore is subjected to fatigue loading. Preventing excessive loading, especially at focal sites, would therefore be necessary to ensure longevity. The use of long constructs (NL) may have resulted in the higher loading rate at the screw-rod interface with stresses of 50 MPa in flexion, 90 MPa in extension, 133 MPa in lateral bending and 131 MPa in rotation, which were much lower than with short constructs. However, the actual restriction may vary among patients as the loading may also vary widely. To determine the clinical relevance, additional studies would be required. The data demonstrate that although the location of peak stress did not change with cross-links, it was reduced at the screw neck and bone-screw interface. Furthermore, due to construct stiffness, the stresses in the vertebra diminish dramatically. This is important since prolonged stress may result in degenerative changes. However, our level of analyses was not rigorous enough to provide conclusive evidence to these findings. Also, our study was conducted using FE analysis with inherent limitations in terms of model creation and data interpretation. Our model includes several assumptions. First, assigning the clinical setting for contact areas between implants and bone is difficult, so the pedicle screw and cage were designed without threads. However, to reduce the model’s inaccuracy, the screws and cage interface with bone was assumed to be perfectly bonded. Second, ligaments with non-linear behavior were assumed, which may have influenced our data. Initially, both linear and non-linear elements were created, but results showed no difference, and hence we assumed linear ligaments. Finally, in designing the annulus fibrosis, fiber layers were not modeled in different layers as is observed in reality. In conclusion, we demonstrated that the use of cross-links in both long and short posterior instrumentation provided better stability by reducing displacement as compared to other configurations. We also demonstrated that the use of this configuration reduced stresses at the screw-bone interface and the dispersal of focal stresses. However, due to the lack of rigor in these analyses, further studies into the stress patterns are required to confirm these findings.

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**REFERENCES**


