EXPERIMENTAL STUDY

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Internal fixation of bilateral sacroiliac dislocation with transiliac locked plate: a biomechanical study on pelvic models

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Objective: The purpose of this study was to analyze and compare the mechanical characteristics of a new iliosacral fixation technique (bilateral S1 pedicle fixation through a transiliac locked plate) for bilateral sacroiliac dislocations with other previously described methods.

Methods: Bilateral sacroiliac dislocations were created in 21 pelvic models and divided into three different fixation method groups. Group 1 was fixed using posterior tension band plating with a 3.5 mm locked plate combined with fixed-angle locked 3.5 mm screw fixation of bilateral S1 vertebra pedicles through suitable holes of the plate. Group 2 underwent posterior tension band plating with a 3.5 mm locked plate combined with bilateral spongious iliosacral screw fixation and Group 3 bilateral iliosacral spongious screw fixation alone. The ultimate load to failure and load for 10 mm of displacement for all three groups were compared.

Results: The average loads to failure for Groups 1, 2 and 3 were 1775, 2084 and 2230 N, respectively, and average loads for 10 mm of displacement were 1033, 1884 and 2013 N, respectively. Group 2 and 3 had the strongest fixation constructs although there was no statistically significant difference between these two groups (p=0.452). Group 2 and 3 were superior to Group 1 in terms of loads for 10 mm of displacement. There was no significant difference between Group 2 and 3 in this regard (p=0.397).

Conclusion: Iliosacral screws are superior to bilateral S1 pedicle fixation through posterior tension band plating. However, the combination of tension band plating with iliosacral screw fixation does not improve the stability of the posterior pelvic ring.

Key words: Iliosacral screw fixation; pelvic ring; sacroiliac dislocation; tension band plating.

Numerous internal fixation techniques and comparative studies on sacroiliac dislocations have been reported in the literature.[1-4] Nevertheless, controversy regarding the best treatment choice remains, particularly for transforaminal sacrum fractures, bilateral sacroiliac dislocations and sacral fractures displaced far laterally. Therefore, biomechanical studies have gained popularity in the last few years.[5]

Several authors have evaluated the biomechanical strength of different internal fixation techniques on the pelvic ring. These biomechanical tests analyzed either the load to failure[6-8] or stability with multiple loading cycles.[5,8,9] These studies differed by origin of bone or bone composite models used, loading conditions and methods to assess fixation stiffness, rendering varied and inconclusive results.[10]

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Fig. 1. Views from the study groups. **(a)** Group 1, **(b)** Group 2, and **(c)** Group 3.

In our study, we aimed to analyze the stability, maximum load bearing capability and failure modes of a new iliosacral fixation method and compare it with iliosacral screws with and without traditional posterior tension band plating in order to analyze mechanical characteristics of a new iliosacral fixation technique.

Materials and methods

The study included 21 artificial pelvises (Model No: 4060; Synbone AG, Malans, Switzerland) specially designed for orthopedic training and scientific research. Models are manufactured from specially formulated polyurethane foam comprising of a cancellous inner core and a harder outer shell simulating cortical bone and have been previously used in several biomechanical studies in the literature.[11-13]

The 21 pelvic models were randomly divided into three fixation groups of 7 models each.

Group 1 received posterior tension band plating using a 3.5 mm locked reconstruction plate with fixed-angle locked 3.5 mm screw fixation of bilateral S1 vertebra pedicles through suitable holes of the plate (Figs. 1a and 2).[14] Group 2 received posterior tension band plating using a 3.5 mm locked reconstruction plate and bilateral 6.5 mm spongious iliosacral screws (Fig. 1b) and Group 3 bilateral 6.5 mm spongious iliosacral screw fixation (Fig. 1c).[15]

A Tile Type C bilateral sacroiliac dislocation was created. For tension band plating, two windows were created 1 cm lateral to the bilateral posterior iliac spines and reconstruction plates contoured, passed through these windows posterior to the sacrum and stabilized with two locked iliac screws apiece. In Group 1, both S1 pedicles were also stabilized with 3.5 mm fixed-angle locked screws passed through the suitable holes of the reconstruction plate.

Vertical stability of each group was determined by axial loading using an automated material testing system (Instron Model No: 4505; Instron Corp., Canton, MA, USA). A specially designed and produced metal interface, stabilizing the model from both of the greater sciatic notches, was used for fixation of the pelvis to the lower jaw of the test machine (Fig. 3). This metal interface is a unique one in the literature and provides a steady experimental model. Load application to the anterior pelvic ring and lateral and anteroposterior rotation and bending of the experimental apparatus was prohibited and a steady, constant and reproducible axial force vector can be applied. For each experiment, the same metal interface was fixed to the lower jaw of the material test machine with a standard configuration. A 20×15 mm bolt connected to the upper mobile jaw applied the vertical load on the midpoint of the S1 vertebra corpus.

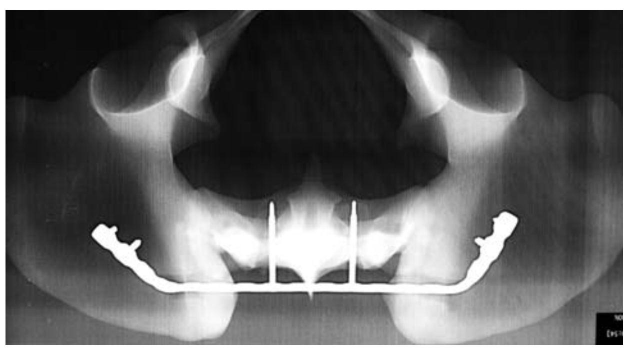

Fig. 2. Radiographic view from Group 1.

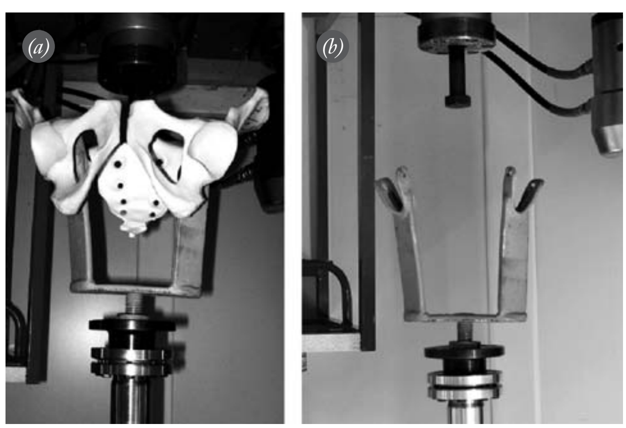

Fig. 3. **(a, b)** Test configuration and material testing machine with the artificial pelvis attached to the jaws.

The distance indicator of the system, which measures changes in the distance between jaws, was set to 0 after application of a precompression load of 50 N. Compression tests were performed at a 5 mm/min rate until failure. Displacement at the sacroiliac joints was measured by the method previously described by Sar and Kilicoglu.[7] Displacement of the Instron crosshead, recorded by the computer, was accepted as the displacement in the sacroiliac joint since the pelvic model was attached to the lower immobile jaw and the sacrum was moving together with the bolt mounted to the upper jaw.[7]

Two different mechanical analyses were performed. The load causing 10 mm of displacement at the sacroiliac joint and the ultimate load of failure were recorded in Newtons.[7] Screw breakage, plate bending, pull-out of the hardware and iliac or sacral fractures were considered as failure.

Mechanical data were recorded simultaneously on a personal computer with a computer-controlled data processor and load to displacement and load to frequency curves of the sacroiliac joints were drawn for all trials.

Statistical analyses were performed using SPSS for Windows v.13.0 (SPSS Inc., Chicago, IL, USA). The Kolmogorov-Smirnov test was used to quantify the normal distribution of the data within each study group and Levene's test to quantify the homogeneity of the variances of each test group.

According to Kolmogorov-Smirnov test, the data had a normal distribution and Levene's test proved homoscedasticity. Thus, analysis of the variance test with post-hoc multiple comparisons was used in order to determine whether there was any significant difference for ultimate load to failures and loads for 10 mm displacement. As the method is an experimental model, the statistical analysis mainly regards this experimental variance between study groups. Significance was set at $p<0.05$.

Results

Average load to failure was 1775±175 N (range: 1550 to 2007 N) in Group 1, 2084±214 N (range: 1740 to 2340 N) in Group 2 and 2230±265 N (range: 1897 to 2606 N) in Group 3. Average load at 10 mm of displacement was 1033±140 N (range: 869 to 1207 N), 1884±163 N (range: 1710 to 2177 N) and 2013±182 N (range: 1815 to 2203 N) in Groups 1, 2 and 3, respectively (Fig. 4). Mean sacroiliac joint displacement was 24.7 (range: 18.2 to 31.1) mm at implant failure in Group 1, 11.7 (range: 6.4 to 15.3) mm in Group 2 and 12.8 (range: 9.84 to 17.1) mm in Group 3. Displacement of 10 mm was recorded before failure in all models in Group 1, in six models in Group 2 and five models in Group 3.

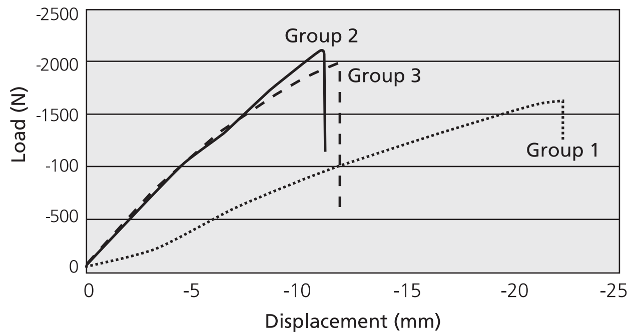

Fig. 4. The mean load to displacement curves of iliosacral joints for each study group. Note that maximum load withstood by Group 1 is less than that withstood by Group 2 and 3.

In Group 1, the most commonly detected implant failure was pull-out of the locked plate from the left iliac wing (4 models), followed by breakage of 3.5 mm screws stabilizing the bilateral S1 pedicles through the holes of the plate (2 models) and pull-out of the hardware from both of the iliac wings (1 model). In Group 2, implant failure occurred at the right sacral ala fracture in 6 models and the left sacral ala fracture in one. Implant failure occurred in the right sacral ala fracture in 5 models and left sacral fracture in 2 in Group 3.

Mean load needed for ultimate load to implant failure increased from Group 1 to Group 3 (Fig. 5a). Between study groups, a right-sided shift of the load-frequency curves was recorded. Although 3 failures occurred in Group 1 between 1550 and 1750N (3/7 pelvises), no failures were noted at the same load interval for Group 3, meaning fixation for Group 3 was relatively stronger than the other groups (Fig. 6a).

Ultimate load to failure was significantly lower in Group 1 than in Groups 2 and 3 (Group 1 vs. Group 2: $p=0.045$; Group 1 vs. Group 3: $p=0.003$). On the other hand, there was no statistically significant difference between Groups 2 and 3 ($p=0.452$).

Mean load needed for 10 mm displacement also increased through Group 1 to Group 3 (Fig. 5B). A right-sided shift of the load frequency curves was recorded between study groups. All 10 mm displacements occurred at the load interval of 750 to 1250 N for Group 1 and 1750 to 2250 N for Group 3. Group 3 required a relatively greater load for 10 mm displacement than Groups 1 and 2 (Fig. 6b).

Loads necessary for 10 mm displacement of Groups 2 and 3 were significantly greater than Group 1 (Group 1 vs. Group 2: $p<0.05$; Group 1 vs. Group 3: $p<0.05$).

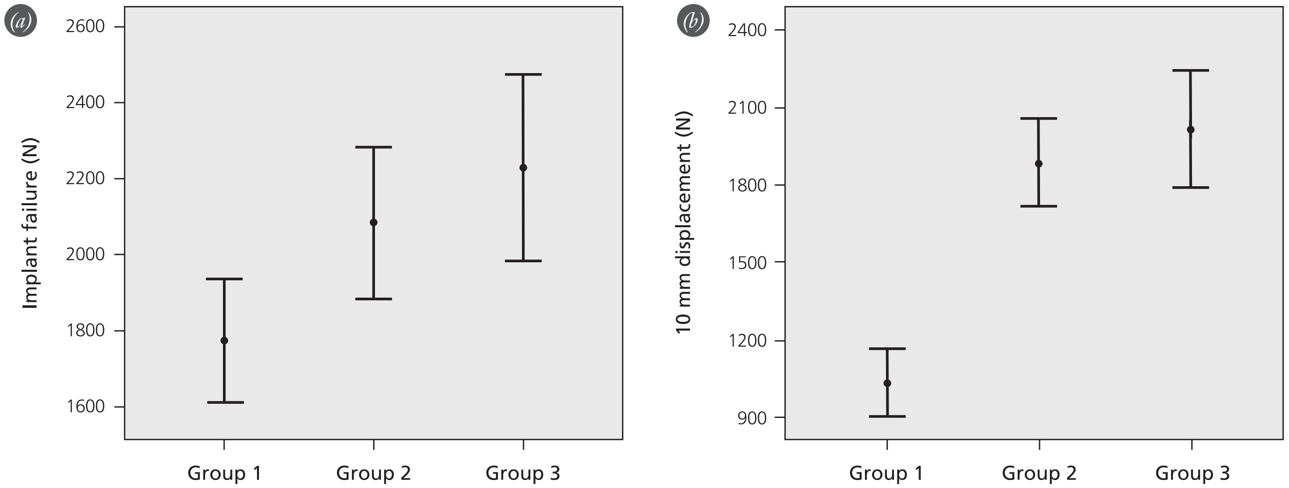

Fig. 5. The mean values and their standard deviations for the study groups. **(a)** Mean values for ultimate load to failure. **(b)** Mean values for 10 mm of displacement. Note the increase in mean values from Group 1 to 3.

On the other hand, there was no statistically significant difference between Groups 2 and 3 (p=0.397).

Discussion

Significant rotational moment is introduced at the posterior pelvic ring with a single or bilateral limb stance as an experimental model.[15,16] Due to the resultant inconsistency and different mechanical outcomes, it is difficult to calculate the exact vectorial forces and maintain its consistency in different loading designs.

In the present study, a specially designed metal interface holding greater sciatic notches bilaterally was used to overcome these disadvantages. As the scope of this current study was not to analyze the biomechanics of the normal pelvis, we did not aim to reproduce the normal force vectors of a human pelvis in vivo. In addition, with the absolute stabilization of the iliums, displacement of the Instron cross-heads was accepted as the vertical displacement in the sacroiliac joints.[7] Therefore an easy-to-repeat experimental model was provided without the necessity for gauges.

In most biomechanical studies, displacement at the fracture site was measured in one or multiple directions.[8,10,11,17-19] In this study, a unidirectional measurement (vertical displacement) was utilized to produce a cost-effective and reproducible experimental modeling and no attempt was made to simulate other pelvic muscles to exclude any unpredictable forces that might influence the measurements.[20]

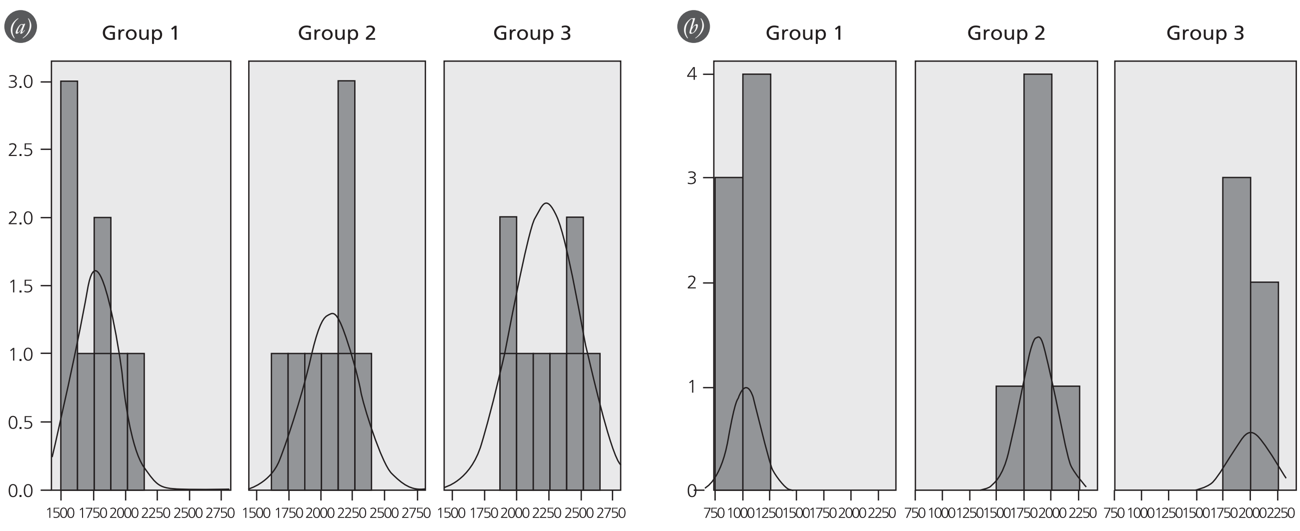

Fig. 6. Load-frequency curves for the study groups. **(a)** Curves for load to failure. **(b)** Curves for 10 mm of displacement. Note the right-sided shift through Group 1 to 3.

Most investigators have used human cadaver specimens for biomechanical analysis.[8,21] Statistical significance is hard to establish due to the difficulty in obtaining fresh cadaver material, variable anatomic configuration and difference in bone stock quality.[5] Other investigators have previously reported the use of cheap and easy-to-obtain composite pelvic bone models in biomechanical studies.[7,10,12,13,20] Although these models are not accepted as an exact analog of the human pelvis, they have some advantages including consistent material properties and minimal variability between study groups.[7,10]

In our study, artificial Synbone pelvises were used. Use of these analogs has been reported previously by other investigators in the literature.[11-13] In a study by Gardner et al.,[11] six Synbones were used successfully in a mechanical study of sacroiliac joint compression. In this study, although unable to stimulate a real human pelvis, Synbones substantially reduced the heterogeneity in size and bone quality associated with cadaveric specimens. The most widely mentioned criticisms about the use of these composite pelvic models depend on their abnormally uniform density throughout the sacrum.[21] Although this criticism is accepted as valid, it is not relevant to our study. In our study, we evaluated the isolated intrinsic biomechanical characteristics of three different fixation methods. Furthermore, as the current study was designed as an experimental study of construct characteristics and biomechanical comparison, any material strength, cortical thickness, or frictional behavior variability between composite bone model and cadaveric specimen was of no concern.

In a sacroiliac joint disruption, sacral bars were found biomechanically inferior to sacroiliac screws and plates.[22] Iliosacral screws and sacroiliac plate gave similar results.[22] The outline of these several biomechanical studies in the literature revealed the repeated use of several different techniques of posterior fixation, including iliosacral screws, anterior sacroiliac joint plates, tension band plates, sacral bars and combined methods.[8,10,16,21]

Yinger et al.[10] reported a biomechanical analysis of 9 different posterior fixation techniques and concluded that two iliosacral screws and a combination of one iliosacral screw and two anterior iliosacral joint plates were consistently the stiffest fixation constructs tested. The data suggest a single iliosacral screw is the least stiff fixation tested. The use of an isolated tension band plate or isolated use of sacral bars was only slightly stiffer than the use of a single iliosacral screw. In our study, our newly described fixation method of posterior tension band plating and S1 vertebra pedicle screw fixation (Group 1) was found mechanically inferior to posterior tension band plating and bilateral spongious iliosacral screw fixation (Group 2) and bilateral spongious iliosacral screw fixation alone (Group 3). There was no statistically significant difference between Groups 2 and 3; a combination of an additional tension band plate with an iliosacral screw fixation does not improve the stability of the posterior pelvic ring. However, in most of the other studies in the literature, there were no biomechanical differences detected between fixation techniques of the pelvis.[8,10]

Traditional posterior tension band plating was first described by Albert et al. and have been evaluated in several studies.[10,14,17,19,23] However, no studies have examined fixation of the S1 vertebra through a posterior tension band plate.[10,19,24] In the classic tension band plating technique described by Albert et al., fixation was achieved with a 4.5 mm conventional plate.[14] However, in the last few decades, less invasive fixation materials and methods have been developed, leading to the more frequent usage of 3.5 mm locked screw plates and their recommendation for the fixation of both anterior and posterior pelvic ring injuries.

In posterior tension band plating, tensile forces on the posterior side of the bilateral sacroiliac joints are converted into compressive forces on the anterior side of the joint. As the plate is placed on the posterior tension side of the joint, with the eccentric load of the pelvis and the sacrum, the compressive forces are obtained from the conversion of the distraction forces by the plate. In the current study, a modification of the traditional tension band plating was defined. First, a 3.5 mm locked reconstruction plate was used in place of a 4.5 mm conventional plate. Second, fixed-angle locked screws were used for plate fixation. The most common implant failure was pull-out of the locked plate from the iliac wings. When these failures were examined, the lack of bone stock of the iliac wings and the low-profile of the locked plate appeared to be the weak points of the construct. Finally, as the plates have fixed-angle locked screws, they had to be inserted high on the iliac wings in order to place the S1 pedicle screws in accurate positions, leading to the bilateral fixation of the iliac wings from the weakest bone stock and high failure rates at the iliac wings. For this reason, we believe that in future studies, better results may occur with the use of a high-profile locked plate (i.e. 4.5 mm locked plate) and fixation of iliac wings with three locked screws instead of two.

Several limitations of this study must be considered. Soft tissue tension or attachments may contribute to the alignment and maintenance of sacroiliac joint reduction.[21] Our research did not include soft tissue factors and may therefore differ from the clinical situation. Second, while our sample size was small, synthetic models with a similar number of specimens have been successfully used in the literature and this number allowed us to demonstrate several significant differences.[7,10,11,20] Another limitation of our study was the usage of the fixed-angle locked 3.5 mm screws for the S1 pedicle fixation. These screws cannot be considered as normal pedicle screws which are thicker and longer than the 3.5 mm locked screws of the plate. Pedicle screw fixation is defined as fixation with screws completely filling the pedicles in the appropriate directions and length. However, such screws were not compatible with the plate used in the current study. Finally, there is currently no clinical use for our novel fixation construct and the magnitude of compression necessary for clinical benefits is unknown. Despite these limitations, we feel that our methodology remains valid for the purpose of this study.

In conclusion, for the fixation of bilateral sacroiliac dislocations, the combination of a tension band plate to an iliosacral screw did not improve the mechanical features and iliosacral screws alone provide a better stability. We believe that the new fixation method described here can lead to future studies and can be used for bilateral sacroiliac dislocations, dislocations with sacral dysmorphism and laterally localized transforaminal sacrum fractures in which iliosacral screw fixation was impossible.

Conflicts of Interest: No conflicts declared.

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