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Evaluation of stabilization methods of pelvic ring injuries by finite element modeling

Pelvis halkası yaralanmaları için kullanılan stabilizasyon yöntemlerinin sonlu elemanlar analizi ile değerlendirilmesi

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Objectives: We developed a finite element model to compare the stabilization methods for the treatment of vertically and rotationally unstable type C pelvic ring injuries.

Materials and methods: Using the COSMOS/M DesignSTAR system, finite element modeling was performed to simulate a type C pelvic ring injury (Denis type 1 fracture of the sacrum and symphysiolysis). The latter was treated with a 4-hole reconstruction plate; the sacrum fracture was fixed with two 2-hole reconstruction plates on the ventral surface or a dorsally applied narrow trans-sacral DC-plate. Finite element modeling included not only the bones and joints, but also mechanically important ligaments. We measured maximum stress in the bones and the ligaments, distribution of stress, displacement and shift. The results were compared with those obtained from measurements performed on cadaver pelvis models.

Results: After direct plating, maximum stress was 21.46 MPa in the plates, and 7.93 MPa in the bones when loading on two feet. These values were 31.1 MPa and 29.1 MPa, respectively when loading on the injured side. With trans-sacral plating, the corresponding figures were 57.36 MPa and 14.14 MPa on two feet, and 223.5 MPa and 23.52 MPa on the injured side. Maximum displacement and shift measured at the two facets of the fracture gap after direct plating were 0.45 mm and 0.01 mm on both feet, and 1.71 mm and 0.28 mm on one foot, respectively. The corresponding figures for trans-sacral plating were 1.099 mm and 0.71 mm on both feet, and 3.55 mm and 2 mm on the injured side. The results of finite element modeling were in agreement with those obtained from bony-ligamentous cadaver pelvis specimens.

Conclusion: Finite element modeling may be useful for the comparison of stabilization methods used for the treatment of pelvic ring injuries.

Key words: Finite element analysis; models, structural; pelvic bones/injuries/physiology; sacrum/injuries; stress, mechanical.

Amaç: Vertikal ve rotasyonel olarak instabil tip C pelvis halkası yaralanmalarının tedavisinde kullanılan stabilizasyon yöntemlerinin karşılaştırılması için sonlu eleman modeli geliştirildi.

Gereç ve yöntem: Çalışmada, COSMOS/M DesignSTAR sistemi kullanılarak hazırlanan sonlu eleman modeli ile tip C pelvis halkası yaralanması (Denis tip 1 sakrum kırığı ve semfizyoliz) oluşturuldu. Semfizyoliz dört delikli rekonstrüksiyon plağı ile tedavi edilirken, sakrum kırığına ventral yüzeyden iki adet iki delikli rekonstrüksiyon plağı veya dorsalden dar bir trans-sakral DC plağı uygulandı. Sonlu eleman modeli sadece kemik ve eklemleri değil, aynı zamanda mekanik olarak önemli ligamanları da kapsıyordu. Bu model ile kemik ve ligamanlardaki maksimum stres, stres dağılımı, deplasman ve şift ölçüldü. Sonuçlar, kadavra modelleri üzerinde yapılan ölçüm sonuçlarıyla karşılaştırıldı.

Bulgular: Doğrudan plak uygulamasında, iki ayak üzerine yük vermede maksimum stres plaklarda 21.46 MPa, kemiklerde 7.93 MPa ölçüldü. Bu değerler, yaralanma tarafına yüklenildiğinde sırasıyla 31.1 MPa ve 29.1 MPa idi. Trans-sakral plak uygulamasında, plak ve kemiklerdeki maksimum stres iki ayak basmada sırasıyla 57.36 MPa ve 14.14 MPa, tek ayak basmada 223.5 MPa ve 23.52 MPa ölçüldü. Kırık aralığının iki yüzünde maksimum deplasman ve şift doğrudan plak uygulamasında iki ayak basmada sırasıyla 0.45 mm ve 0.01 mm, tek ayak basmada 1.71 mm ve 0.28 mm bulundu. Trans-sakral plak uygulamasında ise bu değerler iki ayak basmada 1.099 mm ve 0.71 mm, tek ayak basmada 3.55 mm ve 2 mm ölçüldü. Sonlu eleman modeliyle elde edilen sonuçlar, kadavra pelvis modelleri üzerinde yapılan ölçümlerle uyum göstermekteydi.

Sonuç: Sonlu eleman modeli, pelvis halkası yaralanmalarının tedavisinde kullanılan stabilizasyon yöntemlerinin karşılaştırılmasında yararlı olabilir.

Anahtar sözcükler: Sonlu elemanlar analizi; model, yapısal; pelvis kemikleri/yaralanma/fizyoloji; sakrum/yaralanma; stres, mekanik.

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Operative treatment of vertically unstable, type C pelvic ring injuries is absolutely indicated.[1,2] Considering that these injuries are consequences of high-energy trauma, severe concomitant injuries have to be supposed (polytrauma) on one hand, and on the other hand, soft tissue damage of the hip region is often critical (decolléments). Both issues emphasize the importance of an operative treatment through a minimal exposure, which is less burdensome, easy-to-learn, and quick.

Mears et al.[3] used the "cobra"-plate technique for the treatment of injuries of the dorsal pelvic ring and sacrum. They performed the stabilization of the reduced fracture under direct exposure of the sacrum and the fracture, configuring the plate onto the sacrum and both iliac bones. We investigated whether this technique with some modifications (Fig. 1) would provide appropriate stability and impose less burden to the patient. These modifications make it a semiclosed technique with indirect reduction and extra-focal exposure, whereby the plate is sunk into both iliac bones.[4]

We compared the stability of this modified technique with that of a so called "golden standard" method, where the broken sacrum is stabilized with two 2-hole reconstruction plates inserted from the small pelvis. Comparative stability studies can be performed with bony-ligamentous cadaver pelvis specimens, but such specimens are in short supply, generating numerous technical, sanitary, and ethic difficulties. For this reason, we developed a computer assisted modeling of stability for different surgical techniques. A reliable finite element modeling system would allow to compare stability of other pelvic ring injuries and of different surgical techniques.

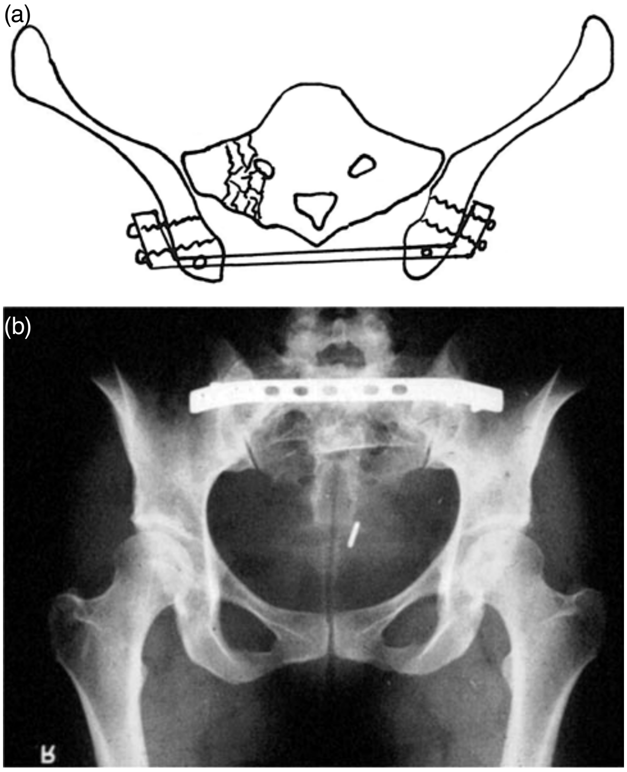

Fig. 1. (a) Schematic delineation of trans-sacral plating and **(b)** its radiographic appearance.

MATERIALS AND METHODS

Anatomical model

We performed parallel experiments on 12 cadaveric bony-ligamentous pelvic specimens (3 females, 9 males; mean age 71.2 years; mean body weight 79 kg). Locomotive disorders/diseases were excluded on the basis of premortem medical histories; no localized morphologic changes could be observed on bony structures. Modeling of vertically unstable injury was performed as follows: artificial symphysiolysis was created in the ventral pelvic ring and was stabilized with a 4-hole narrow DC-plate (Fig. 2). Dorsal ring injury in the lateral mass of the sacrum was first fixed with two 2-hole reconstruction plates applied directly through a ventral approach, and then was stabilized with a 10-12-hole narrow DC-plate from a dorsal exposure (so called "trans-sacral plate"). Keeping the femur on the injured side fixed, a physiological load of 250 N was exerted onto the sacral promontory. The displacement between the two facets of the fracture gap was recorded in each specimen.

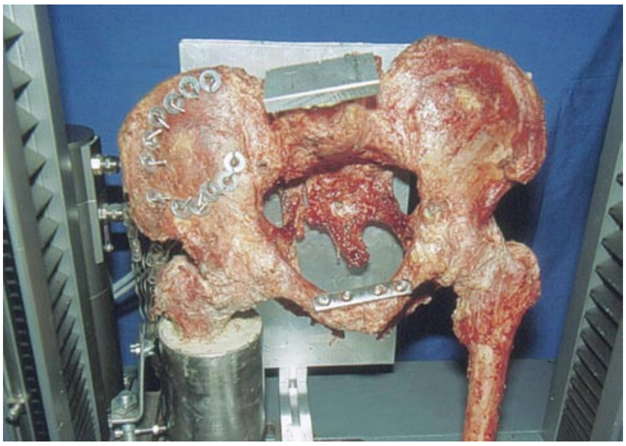

Fig. 2. Model of the injury demonstrated on a bony-ligamentous cadaver pelvis specimen.

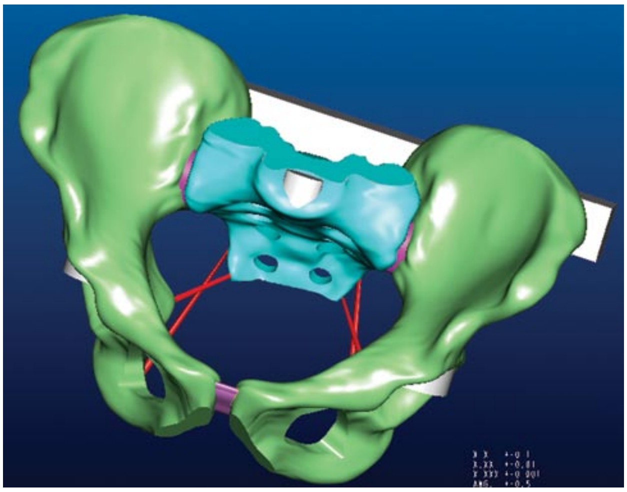

Fig. 3. Scanned image of a plastic pelvis.

Geometric model

There were two options: *(i)* creating a model on the basis of computed tomography (CT)-scans requiring special software; *(ii)* scanning a plastic model using a three-dimensional measuring device (Fig 3). The second option was favored by the fact that connection with the computer of the CT required special software, and the model created in this way basically would not be compatible with the software developed for technical purposes. Furthermore, the special software would demonstrate all bones of the pelvis as a uniform unit, and modeling of the joints among them would result in additional difficulties.

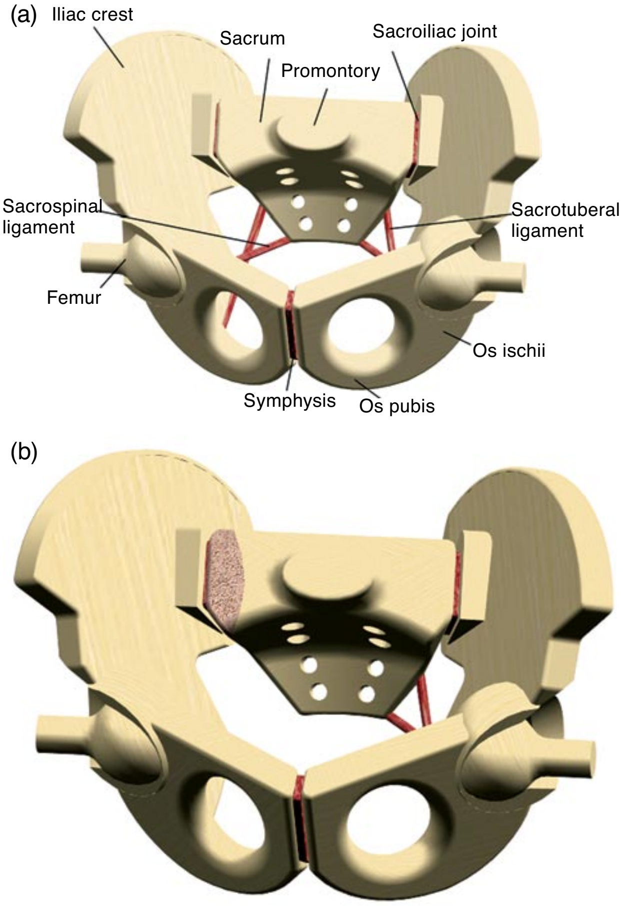

Fig. 4. Simplified geometrical model: **(a)** intact pelvis; **(b)** broken sacrum where the sacrotuberal and sacrospinal ligaments are missing on the injured side because of disruption caused by the presumed pelvic injury.

Due to the sophisticated geometry of the pelvis, finite element program was not applicable for creating a mesh of the model. That is why we developed a "pelvis-like" geometric body consisting of simplified geometric objects (Fig. 3).[5] "Node-to-node" contact junction was modeled in the hip joints and "bonded" junction in the intact, tight joints (symphysis and sacroiliac joint) (Fig. 4a). Fracture pattern of the sacrum is demonstrated in Fig. 4b. There was a "node-to-node" contact junction between the two facets of the fracture gap in the broken pelvis and the two fragments of the injured symphysis. Originally, we fixed the plates by gluing the bolt´s head at the screw holes. In the course of model development, we succeeded in demonstrating the screws as cylinders (Fig. 5).

Material of the model

We considered the anatomical units of the pelvic ring to be a linearly elastic, isotropic material. In a

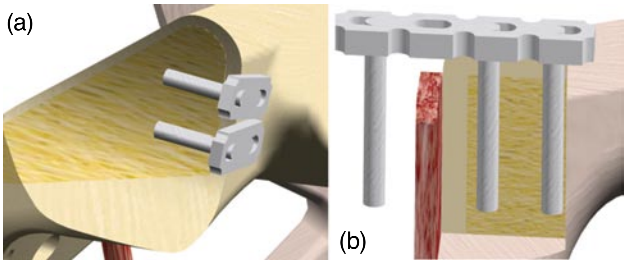

Fig. 5. Separation of cortical and cancellous bone adjacent to the screws: **(a)** sacrum, **(b)** symphysis.

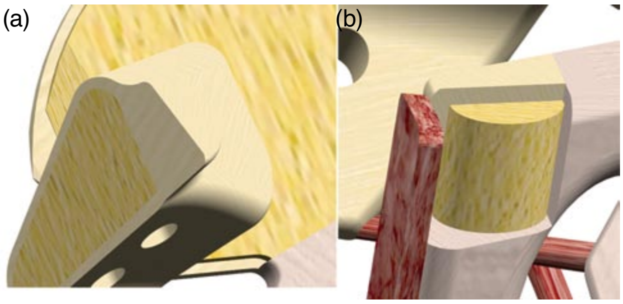

Fig. 6. **(a)** Perpendicular section of the sacrum; **(b)** separated cortical and cancellous bone of the symphysis.

TABLE I

Material properties of bones, joints, ligaments, and plates[6]

	Elastic modulus	Poisson's ratio
Bones		
Cortical bone	17,000 MPa	0.3
Cancellous bone	400 MPa	0.2
In the ratio of their volume (10% cortical bone+ 90% cancellous bone)	2,060 MPa	0.21
Joints		
Sacroiliac joint	68 MPa	0.2
Symphysis	50 MPa	0.2
Ligaments		
Sacrospinal ligament	355 MPa	0.2
Sacrotuberal ligament	355 MPa	0.2
Plates		
DC-plate	200,000 MPa	0.28
Reconstruction plate	200,000 MPa	0.28
Maximal allowable stress values		
Cortical bone	70 MPa	
Cancellous bone	15 MPa	
Plates	800 MPa	

previously developed model, we could not differentiate the cortical and cancellous bone substance, because at this time the finite element software was not suitable for scanning the model.

In the course of development, we could model the distinct 3-mm thick cortical and accordingly thick cancellous bone substance at crucial points (in the front, at the symphysis, and dorsally in the broken sacrum, where stabilization would be performed) (Fig. 6). We applied material characteristics published in the work of Abé et al.,[6] who built up their work using material characteristics of living tissues. Accordingly, in the course of model construction we used material parameters listed in Table I.

Loading, boundary conditions

Boundary conditions were defined so that they coincided with those obtained from stability investigations performed on cadaver pelvis specimens. Thus, the results of investigations performed in cadavers could be compared with those of the computerized model. Six cases were chosen for finite element analysis.

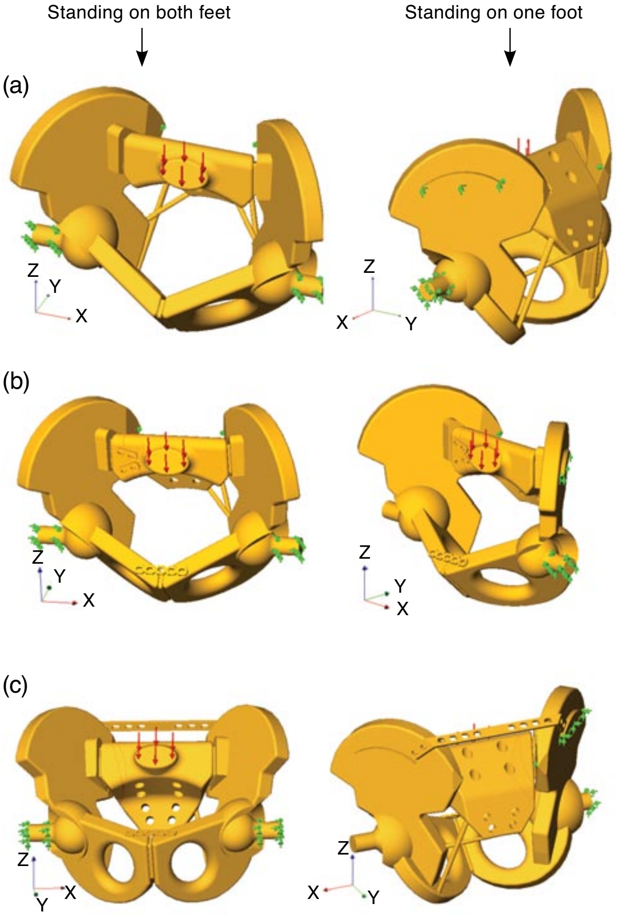

Fig. 7. (a) Intact pelvis; **(b)** direct plating; **(c)** trans-sacral plating.

Case 1 and 2

Intact sacrum, standing on one foot and two feet, respectively. Load on the sacral promontory was exerted in direction Z at a force of 500 N, with one femur kept fixed, pelvis dorsally supported against shift into direction Y, "node-node" type junction in both hip joints, and "bonded" junction on other surfaces (Fig. 7a).

Case 3 and 4

Broken sacrum, direct plating, and standing on one foot and two feet, respectively. Broken surfaces and the symphysis were joined in a "node-node" manner, all the rest as in case 1 (Fig. 7b).

Case 5 and 6

Loading and boundary conditions and the type of junctions between the corresponding parts of the pelvis were identical to those of trans-sacral plating and direct plating (Fig. 7c).

Using the COSMOS/M DesignSTAR system (ver. 4.0, Structural Research and Analysis Corporation,

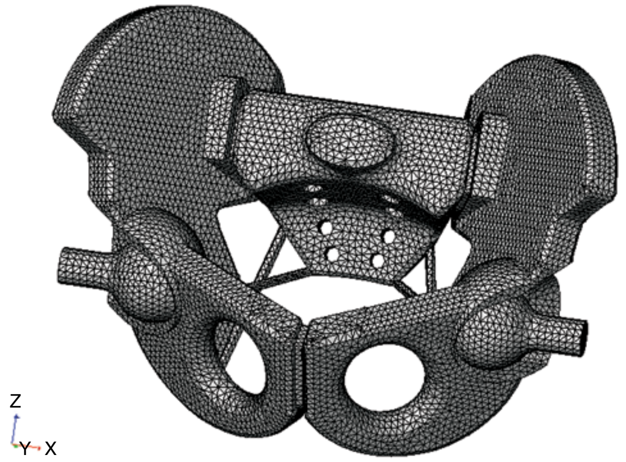

Fig. 8. Finite element screening.

Santa Barbara, CA, USA)[7] we applied 4-node tetrahedron elements with an average element size of 4 mm and achieved a finite element model consisting of 172,100 elements and 36,000 nodes (Fig. 8).

RESULTS

Case 1

In case of intact pelvic ring and standing on both feet, the maximum of the equivalent stress was 5.19 MPa in the cortical bone of the sacrum and 2.73 MPa at the origin of the sacrospinal ligament. The distribution map of the developing arch-like tension met our expectancies. The maximum displacement was 0.318 mm in X and Y directions and the sacrum tilted - as expected - around an axle running through the sacroiliac joint (Fig. 9).

Case 2

In case of standing on one foot, the maximum of the equivalent stress was 6.01 MPa in the bones and 3.49 MPa in the ligaments. The arch-like distribution of the stress changed; higher stress arose in the anterior part of the pelvic ring compared to that occurred when standing on both feet. The maximum displacement was 1.41 mm; the hipbone on the supported side remained fixed due to its stabilization; displacement constantly increased

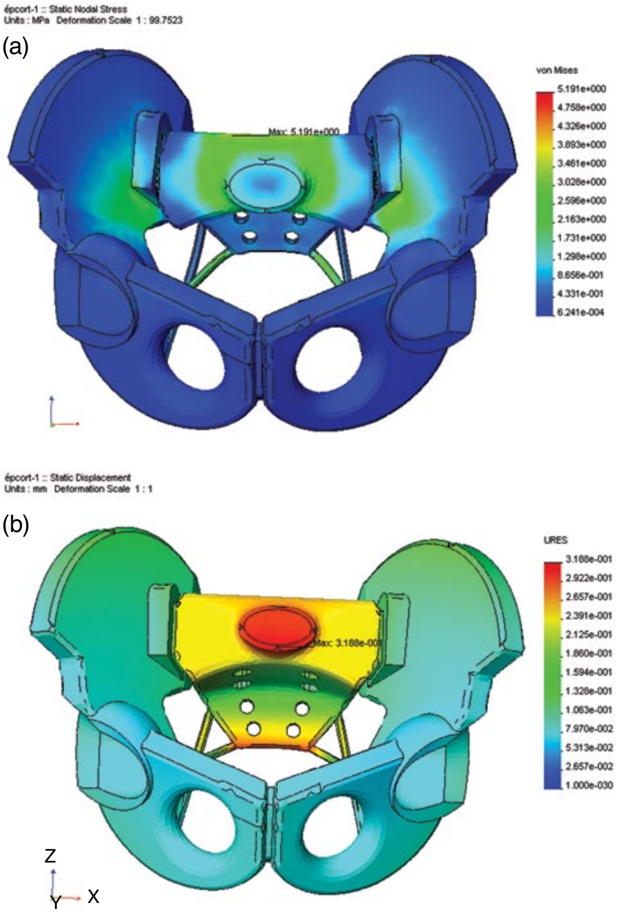

Fig. 9. (a) Tension and **(b)** shift in case of intact pelvis.

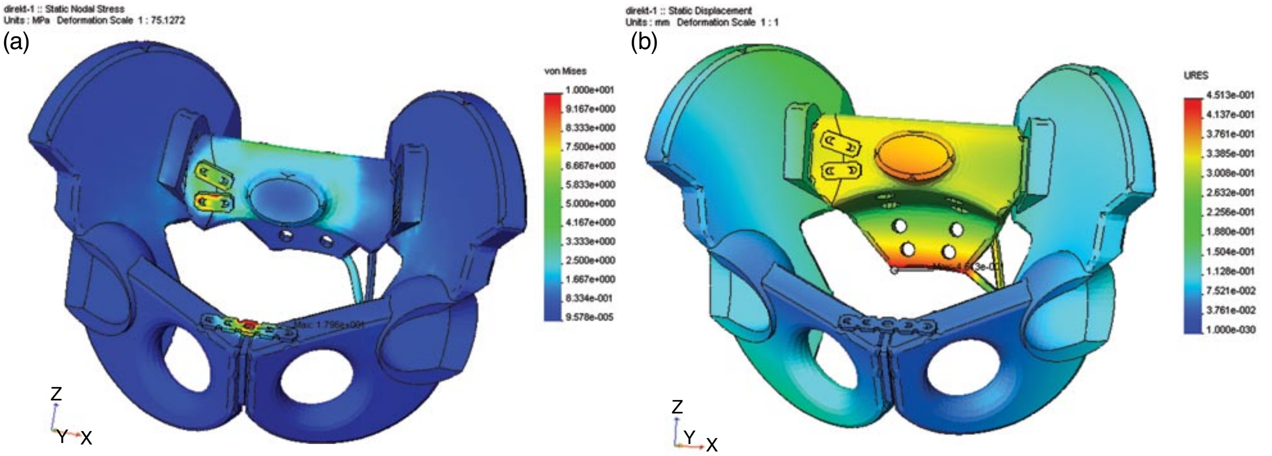

Fig. 10. (a) Tension and **(b)** shift in case of direct plating and standing on both feet.

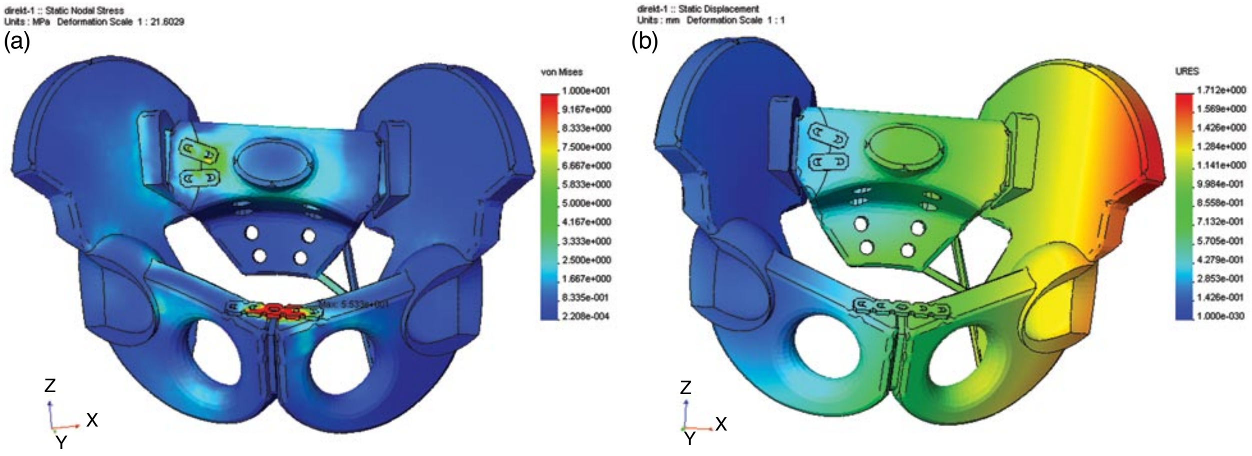

Fig. 11. (a) Tension and **(b)** shift in case of direct plating and standing on the foot on the injured side.

toward the unsupported side. It was concluded that our model met the expectancies in terms of anatomical precognitions and, thus, was suitable for further analysis.

Case 3

After direct plating, in case of standing on both feet, the arch-like distribution of the equivalent stress and its values remained unchanged. Equivalent stress was 21.46 MPa in the plates, and 7.93 MPa adjacent to them and in the bones. Compared with our initial model, the peak of the stress decreased due to a more lifelike modeling of the position of the screws in the bone. Stresses smaller in the anterior plate, likewise in the anterior part, is associated with smaller stress in the intact pelvis, as well. Maximum displacement was 0.45 mm, and the shift in the fracture gap was 0.01 mm (Fig. 10).

Case 4

In case of standing on the foot on the injured side, stresses increased in the anterior plate due to exertion of higher load in the anterior part of the pelvic ring. Maximum stress was 31.1 MPa in the plates, and 29.1 MPa in the bones. The shift pattern was similar to that of the intact pelvis. Maximum displacement was 1.71 mm, and the shift in the fracture gap was 0.28 mm (Fig. 11).

Case 5

Dorsal plate enabled power transfer in case of trans-sacral plating. In case of standing on both feet, a stress of 57.36 MPa occurred in the plate and 14.14 MPa in the bones. Maximum displacement was 1.099 mm, and the shift in the fracture gap was 0.71 mm (Fig. 12).

Case 6

Loading the extremity on the injured side, the maximum stress was 223.5 MPa in the plates, and 23.52 MPa in the bones. Maximum displacement was 3.55 mm, and the shift in the fracture gap was 2 mm. Shift values in Case 5 and 6 were measured in a model with smooth fracture surfaces.

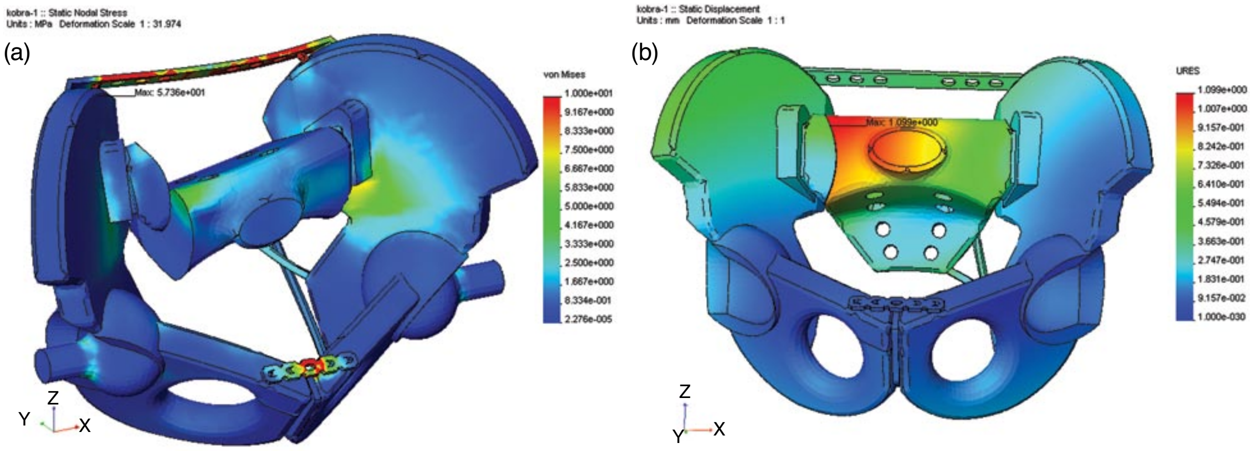

Fig. 12. Trans-sacral fixation: **(a)** Tension and **(b)** shift in case of standing on both feet.

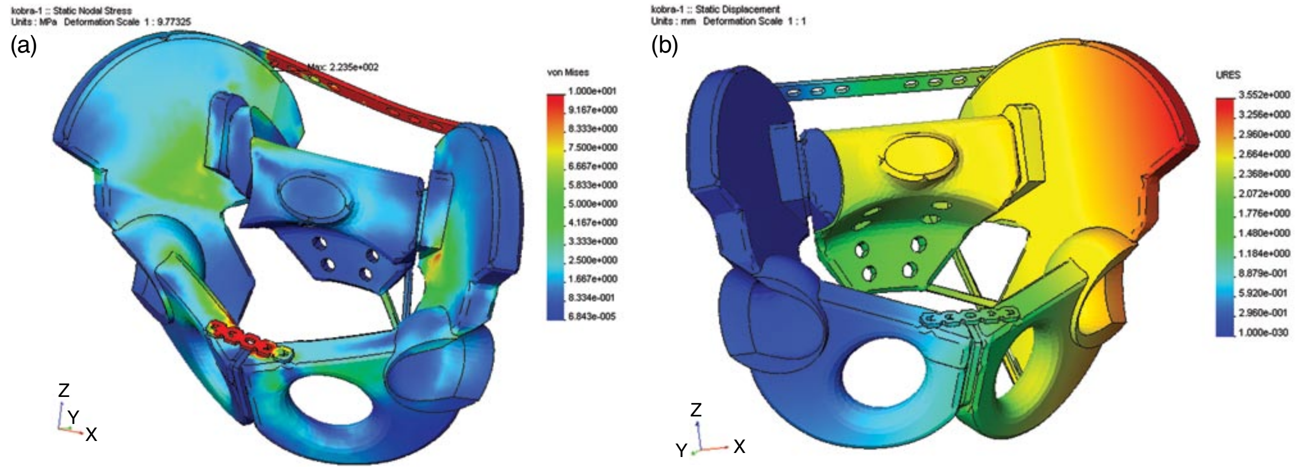

Fig. 13. Trans-sacral fixation: **(a)** Tension and **(b)** shift in case of standing on the foot on the injured side.

Compared with real cases, these values were considerably higher, since in case of uneven fracture surface the shift would be appreciably smaller due to friction (Fig. 13).

DISCUSSION

In our study, we compared two operative ways of stabilization. “Direct” plating provides more stability, but it is considerably more burdensome to the patient. Trans-sacral plating performed in a semiclosed manner is less invasive, but provides less stability.

Ethical, sanitary, and technical difficulties in measuring cadaveric bony-ligamentous pelvis specimens impose considerable limitations on the availability of experimental results. Thus, computerized, finite element modeling of bony-ligamentous specimens is increasingly preferred. Finite element modeling of the intact pelvis was published previously.[8] In our study, we developed a comparative modeling of stability of two operative ways of stabilization in case of a certain type of injury on the basis of results obtained from experi-

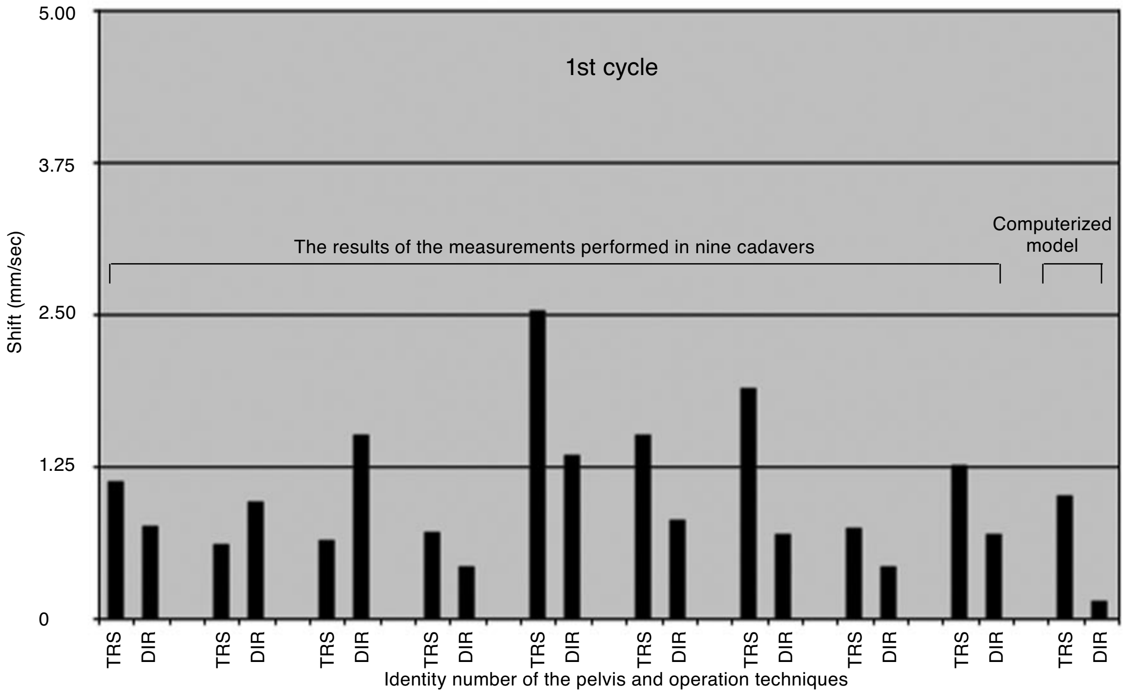

Fig. 14. Shift values (mm) between the two facets of the fracture gap in cadaveric bony-ligamentous specimens and in the computerized model after trans-sacral plating (TRS) and direct plating (DIR) with a load of 250 N, standing on the foot of the injured side.

ments on cadaveric specimens. Naturally, finite element modeling provides only approximate results compared with reality, so the following shortcomings have to be considered: *(i)* Fracture surfaces are not completely smooth, their friction coefficient is unknown. *(ii)* The geometric model of the pelvis is only approximate, material characteristics applied on this model are not uniform in the literature. *(iii)* We also investigated the situation in full weight-bearing of the injured side, which is not allowed during the first 6 to 8 weeks.

In view of stress values obtained from the neighboring of the screws, we may also conclude that the conjunction is stable and the screws do not cut out. Since stresses occurring in the plates do not exceed allowable limits, the plates sustain the load without permanent deformation. Regarding the load, direct plating provides - as expected - higher stability; however, our clinical and radiological follow-up observations suggest that trans-sacral plating provides sufficient stability for mobilization without weight-bearing of the injured side and for bone healing.

In the course of parallel experiments on cadaveric bony-ligamentous specimens, 12 pelvises were at our disposal and we could obtain utilizable results from nine of them. We recorded the displacement between the two facets of the fracture gap in each specimen; first after “direct” plating, and, subsequently, after trans-sacral plating. On the basis of these experimental results (Fig. 14), we may conclude that the results of the finite element modeling can be used for a good estimation. Accuracy of the model can be improved by considering its approximations described above.

Acknowledgments

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