Finite Element Analysis and Experimental Testing of Stiffness of the Sarafix External Fixator

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Abstract

This paper presents research results of a stiffness analysis of a Sarafix external fixation system. The stiffness property of the external fixator affects the local biomechanical and biological environment of fracture healing. The research has been conducted in a case when one of the Sarafix unilateral biplanar fixator configurations has been applied to a tibia with an open fracture. A stiffness analysis was performed using FEA and experimental testing under three types of loads: axial compression, AP (anterior-posterior) four-point bending and torsion. 3D geometrical and FEM model of the fixator configuration was formed using CATIA V5 software system and a structural analysis was performed afterwards. Verification of the results obtained from a structural analysis was carried out through experimental testing by comparing values of an appropriate component of displacement at the point of load.

Keywords: the Sarafix system; finite element method; experimental testing; displacements; stiffness

1. Main text

After J.F. Malgaigne invented the fixator in 1840, their selection and application was generally carried out on empirical grounds and accumulated experience in clinical orthopaedics and traumatology. In order to promote and carry out necessary research to improve fixation, a development of a theoretical analysis of problems fixation based on the principles of structural mechanics is pursued.

The external fixator is a medical device for the immobilization of fractures or serious damage to the structure of extremities. External fixation is a method of fracture immobilization achieved by the application of pins or wires

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into or through a bone and their binding to the outer frame [2].

The above basic concept of the method has not changed since its origin, but progress is reflected through the development of new design solutions and materials used. In the last two decades, a closer link between medical science and other disciplines of science (Technics, Medical Engineering, Biomechanics etc.) has been created, with the aim of multidisciplinary solving contemporary medical problems. One example of association of scientists of different profiles for the purpose of designing and improving medical equipment is the application of methods of external fixation and the development of systems for external fixation.

The idea for the development of the external fixator Sarafix was developed by a group of orthopaedists of "prim.dr. Abdulah Nakas" General Hospital in Sarajevo under siege, in May 1992 [1]. The idea was triggered by the insufficient number of existing fixators, as the result of the expansion of the war activities. Shortly after, the first fixator called Sarajevo war fixator - Sarafix was produced (Fig. 1.).

During the war, the Sarafix found its highest application in the treatment of extensive gunshot-explosive fractures of long bones of the extremities. Today, in peacetime traumatology, it is used in accidental injury in traffic accidents and industrial trauma. Sarafix external fixation system represents a unilateral, biplanar external fixator which belongs to a group of modular fixators with one-half pins. Owing to the high flexibility and mobility, its application is possible to the complete human skeleton. Sarafix is the holder of numerous awards and prizes at international exhibitions of innovations, and gold medals at the exhibitions of innovations Brussels Eureka 95 and Geneva 1996, and Sarajevo's Sixth of April Award for 2001 should be emphasized [1].

2. Objective and Methods

All commercial fixators, now in use, passed a biomechanical study before their first application. Mechanical testing of Sarafix fixator was not performed before its clinical application, because of the war-time circumstances in which it originated. Complete mechanical research of the fixator, besides the examination of its stiffness to the loads to which it was exposed after the application, includes the analysis of stresses (von Mises and principal stresses) on the characteristic location of fixator design. Extensive studies of the mechanical research of the Sarafix fixator were carried out within the thesis [5]. Due to the limited scope of this paper, only the results of the stiffness analysis will be presented.

The main value for evaluation of mechanical stability of the external fixator is fixator stiffness. With the aim of determining stability of external fixators, various sensors and transducers are set up on their designs [2]. During the past few years, except of performing the experimental testing, there has been an increased use of geometrical modeling and finite element analysis (FEA), in order to more fully describe the behaviour of the fixator and its components during the loading [4, 9].
One of the reasons for determining stiffness of the external fixators is its impact on the stress generated in the contact of one-half pin-bone. Increasing the stiffness of the fixation device significantly reduces the axial load to the one-half pins, and thus stresses generated at the one-half pin-bone contact. This helps reduce the risk of weakening (relaxation) of the one-half pins and infection in the area around the one-half pin, which is usually related to complications of external fixation of bone [1, 2]. On the other hand, after the initial phase of treatment, for the purpose of dynamization process and in order to stimulate consolidation of the bone, it is desirable to control fixator stiffness and coordinate to the trend of fracture healing.

This paper presents results of stiffness analysis of the most used configuration of the Sarafix external fixator in the case of an unstable tibial fracture. An open fracture at the middle of tibia with fracture gap of 50 mm (severe extensive injury with a considerable defect of bone structure) was examined. The most complicated aspect of bone fractures, both in terms of complexity of treatment and structural stresses of external fixator, is an open fracture. In the case of open fractures, in the initial phase of treatment, the full load is transferred through the fixator.

The analyzed configuration of the Sarafix fixator contains four one-half pins in proximal and distal bone segment as shown in Fig. 2 and 3. The stiffness analysis of the Sarafix fixator was carried out using FEA and experimental analysis under three types of loads: axial compression, AP (anterior-posterior) four-point bending and torsion. Understanding the physical behaviour of the model is a basic prerequisite for successful process of modeling real systems. Before that, it is necessary to make numerous assumptions related to modeling: structure, joints between the components, boundary conditions, loads and materials.

Geometrical modeling of the Sarafix fixator and FEA were carried out at the Laboratory for Computer Aided Design - CADlab of the Faculty of Mechanical Engineering Sarajevo. The first step consisted of forming a 3D geometrical model of the analyzed Sarafix fixator configuration, whereupon the FEA was performed on the model using CAD/CAM/CAE (Computer Aided Design/Computer Aided Manufacturing/Computer Aided Engineering) system CATIA. Figure 2 shows the CAD and finite element method (FEM) model of the analyzed Sarafix fixator configuration after pre-processing. During the processes of the linear FEA, the material of wooden bone models was defined as orthotropic, while materials of the fixator were modeled as isotropic. The FEM model consisted of solid finite elements of a linear (TE4) and parabolic tetrahedron (TE10) type. Join elements of the spider type were used for modeling the joints between the components of the Sarafix fixator. The following joints were used: Fastened connection, Contact connection and Bolt tightening connection [5, 7].

![Fig. 2. 3D CAD and FEM model (axial compression) of the analyzed Sarafix fixator configuration.](image)
Experimental testing was conducted at the Laboratory for materials testing and Laboratory for production engineering of the Faculty of Mechanical Engineering Sarajevo as shown in Fig. 3. At the Laboratory for materials testing, the examination of the analyzed configuration of the Sarafix fixator on the axial compression and AP bending was performed, using a universal material testing machine (Zwick GmbH & Co., Ulm, Germany, model 143501). The analyzed configuration of the Sarafix fixator was attached to proximal and distal tibia bone segments modeled with cylindrical wooden bars with known physical properties. During the testing, the intensity of the load (0 to 600 N at the rate of 5 N/s) on the model of proximal segment of the tibia was controlled, using the force transducer (U2A, HBM-Hottinger Baldwin Messtechnik GmbH, Darmstadt, Germany). A wooden model of the proximal and distal bone segments are supported on the ball joint supports [5, 6].

Most biomechanical studies of the external fixation analyze only total characteristics of stiffness of diverse types of fixators and configurations [2, 4, 9]. This paper, except the value of the Sarafix fixator stiffness, analyzes also fracture stiffness. One of the possibilities of structural analysis using FEM is to determine the direction and intensity of movement of any point of the bone models and fixator.

Fixator stiffness is an important mechanical characteristic, but it cannot provide direct information about displacement of a fracture gap. The precise information can be provided by analyzing relative displacements of end bone segments under simulated conditions of loads. However, in addition to numerous research, it remains unclear which forms of movement are helpful and harmful to the healing of fractures, therefore the information about the values of relative movement of the bone parts is of limited value. But on the basis of literature the following two hypotheses [9] could be suggested:

- Cyclic axial micro motion is beneficial for healing of fractures
- Shearing motions of bone segments at the fracture site are detrimental to its healing

Relative craniocaudal and lateromedial displacements (x and y direction) and axial displacements (z direction) for analyzed points were calculated as:

\[ r_{D(x)} = D_{p(x)} - D_{d(x)}; \quad r_{D(y)} = D_{p(y)} - D_{d(y)}; \quad r_{D(z)} = D_{p(z)} - D_{d(z)} \]  \( (1) \)

where:

- \( r_{D(x)}, r_{D(y)} \) and \( r_{D(z)} \) are the relative displacements at the fracture gap in the x, y and z directions (mm),
- \( D_{p(x)}, D_{p(y)} \) and \( D_{p(z)} \) are the absolute displacements of points at the proximal fracture endplate in the x, y and z direction (mm),
$D_{d(x)}, D_{d(y)}$ and $D_{d(z)}$ - are the absolute displacements of points at the distal fracture endplate in the x, y and z direction (mm).

Based on the values of relative displacements $r_D$, maximal value of the resulting vector of relative displacements at the fracture gap (under the loads) is determined as:

$$R = \sqrt{(r_{D(x)})^2 + (r_{D(y)})^2 + (r_{D(z)})^2}$$

(2)

Absolute displacements of analyzing points at the proximal and distal fracture endplate in the x, y and z direction were determined. Analyzing points were selected in such a manner for the resulting vector of relative displacements ($R$) has maximal value.

3. Stiffness analysis on axial compression

During the axial compression testing, the bone models were supported on ball joints, while maximal axial loading force applied to the proximal bone model was $F_p = 600$ N. The modeling of the influence of supports was performed using a Smooth virtual part. At the end of the proximal bone segment, the axial load in the form of surface force (Force density) was applied in the direction of the z axis of the Cartesian coordinate system. A displacement constraint of the Sarafix FEM model was derived by using the Ball join restraint on the model of distal bone segment. Likewise, a displacement constraint at the model of proximal bone segment was performed by using the User-defined restraint, which prevented the two translations in direction of x and y axis of the Cartesian coordinate system as shown in Fig. 2 and 3.

Figure 4 shows the 3D FEM model of the analyzed configuration Sarafix fixator before and after the action of maximum axial load. The directions and intensities of deformation of each point of the structure of the system and bone models are observed in Fig. 5.

Axial fixator stiffness ($C_p$) was calculated using the following equation:

$$C_p = \frac{F_p}{\delta_p} \text{ (N/mm)}$$

(3)

where:
$F_p$ – is the applied axial loading force (N),
$\delta_p$ – is the axial displacement of proximal segment at the point of load (mm).

Diagram of axial displacement proximal segment model of bone at the point of load, as shown in Fig. 5, was obtained by the structural analysis using FEM and experimental testing. It shows the intensity of deformation of the analyzed Sarafix fixator configuration during testing under axial compression.

![Graph](image)

Fig. 5. Comparative diagram of the axial displacement at the point of load.

Values of bone segments displacements at the point of load and fracture gap have been analyzed. Verification of the results obtained from a FEA was carried out through experimental testing by comparing values of an appropriate component of displacement at the point of load ($\delta_p$). The axial fracture stiffness was calculated as the applied axial force divided by total displacement at the analyzing points [5, 8]:

$$C_{pp} = \frac{F_p}{R} = \frac{F_p}{\sqrt{(r_{D(x)})^2 + (r_{D(y)})^2 + (r_{D(z)})^2}}$$

(4)

4. Stiffness analysis on AP bending

During the testing under AP four-point bending models of the bones are free to rely on the cylinder with a flat surface reliance. Maximum value of bending force was: $F_s = 500$ N. Figure 6 shows the experimental set-up for testing on AP bending.

![Image](image)

Fig. 6. Experimental testing of Sarafix fixator on AP bending.

Figure 7 shows the 3D FEM model of the analyzed Sarafix fixator configuration before and after the action of
maximum bending force, which acts simultaneously on both models of bone segments. Directions and intensities of
displacement of every point of the system structure and bone models are noted clearly.

Fig. 7. Non-deformed/deformed structure of the Sarafix fixator and translation displacement vectors at the fracture gap under maximum bending force.

AP bending fixator stiffness \( (C_s) \) determined as follows:

\[
C_s = \frac{F}{\delta_s} \quad \text{(N/mm)}
\]  \hspace{1cm} (5)

where:

\( F \) – is the applied bending force (N),

\( \delta_s \) – is the displacement (deflection) of bone segment at the point of load (mm).

Verification of the results obtained from a FEA was carried out through experimental testing by comparing
values of deflection at the point of load (\( \delta_s \)). Diagram of the displacement proximal and distal segments of the bone
models at the point of load was obtained by structural analysis using FEM and experimental testing as shown in Fig.
8. It shows the intensity of deformation of the analyzed Sarafix fixator configuration during testing under AP bending.

Fig. 8. Comparative diagram of deflection at the point of load.

The fracture bending stiffness was calculated as the applied bending force divided by total displacement at the
analyzing points [5, 8]:

\[
C_{ps} = \frac{F}{R} = \frac{F}{\sqrt{(r_{D(x)})^2 + (r_{D(y)})^2 + (r_{D(z)})^2}}
\]  \hspace{1cm} (6)
5. Stiffness analysis on torsion

Testing under torsion was carried out by the principle of rotation distal segment of the bone model in relation to the fixed proximal segment. At the end of the upper segment of bone model, in the appropriate place we put constraint in the form of a cylindrical bearing as shown in Fig. 9.

![Fig. 9.](image)

Torsion moment is defined on the whole surface in the segment of bone model. The maximum value of the torque was: $M_u = 15$ Nm.

![Fig. 10.](image)

Torsion fixator stiffness ($C_u$) is calculated using the following relation:

$$C_u = \frac{M_u}{\theta} \text{ (Nm/rad)} \tag{7}$$

where:

- $M_u$ – is torque (Nm),
- $\theta$ – is torsion angle of bone segment at the point of load (rad).

Figure 10 shows the 3D FEM model of the Sarafix fixator configuration during structural analysis under torsion. Rotation of the system structure points after acting of the maximum torque is noted clearly. Using structural analysis and experimental testing a diagram of torsion angle of proximal segment model of bone at the point of load was obtained. Figure 11 shows this scenario. It also shows the intensity of deformation of the Sarafix fixator configuration during testing under torsion.

![Fig. 11.](image)
The fracture torsion stiffness was calculated as the applied torque divided by total displacement at the analyzing points [5, 8]:

\[ C_{pu} = \frac{M_u}{R} = \frac{M_u}{\sqrt{(r_{D(x)})^2 + (r_{D(y)})^2 + (r_{D(z)})^2}} \]  

(8)

6. Results

According to the given geometrical configurations, acquired values of displacement of proximal and distal model of the bone segment at the fracture gap under maximal axial load, bending force and torque are presented in Table 1.

Displacements were analyzed at the point of load and fracture gap using FEM and experimental testing. Based on the displacement at the point of load (\( \delta \) i \( \theta \)), the values of the fixator stiffness (C) are determined, based on the relative displacements at the fracture gap (R), the values of fracture stiffness (Cp) are determined as shown in the Table 1.

Table 1. Values of stiffness and displacements under maximum intensity of loads.

<table>
<thead>
<tr>
<th>Type of load</th>
<th>Analyzing method</th>
<th>Displacement of the proximal segment at the fracture gap (mm)</th>
<th>Displacement of the distal segment at the fracture gap (mm)</th>
<th>Maximum relative displacement at the gap (mm)</th>
<th>Displacement at the point of load (mm; rad)</th>
<th>Fracture stiffness (N/mm; Nm/mm)</th>
<th>Fixator stiffness (N/mm; Nm/rad)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Axial compression</td>
<td>FEM</td>
<td>0,53 4,14 -4,36</td>
<td>0,53 4,29 0,22</td>
<td>4,58</td>
<td>4,18</td>
<td>130,93</td>
<td>143,54</td>
</tr>
<tr>
<td></td>
<td>Exp.</td>
<td>- - - -</td>
<td>- - -</td>
<td>-</td>
<td>-</td>
<td>4,35</td>
<td>-</td>
</tr>
<tr>
<td>AP bending</td>
<td>FEM</td>
<td>-0,03 2,8 -0,53</td>
<td>-0,03 2,8 2,21</td>
<td>2,74</td>
<td>2,48</td>
<td>182,22</td>
<td>201,61</td>
</tr>
<tr>
<td></td>
<td>Exp.</td>
<td>- - - -</td>
<td>- - -</td>
<td>-</td>
<td>-</td>
<td>2,59</td>
<td>-</td>
</tr>
<tr>
<td>Torsion</td>
<td>FEM</td>
<td>0,82 0 0,05 0 0</td>
<td>0,76 0,073 19,74</td>
<td>205,48</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>Exp.</td>
<td>- - - -</td>
<td>- - -</td>
<td>-</td>
<td>-</td>
<td>0,076</td>
<td>-</td>
</tr>
</tbody>
</table>

It is known that the directions and intensities of interfragmentary displacements in fracture gap (\( D_p \) and \( D_d \)), as well as stiffness of external fixator (C), affect the outcome and speed of the fractures’ healing. Interfragmentary displacements parallel to the fracture surfaces (\( D_{p(y)} \) and \( D_{d(y)} \)), lead to the appearance of pseudo-arthrosis instead of fracture healing. For these reasons, it is necessary to control interfragmentary displacements, especially to minimize transverse (shearing) displacements of bone ends at the fracture gap (\( D_{p(y)} \) and \( D_{d(y)} \)).

Conclusion

The conducted research has shown that there is a linear dependence between the load and displacement of the bone segments, as a result of the absence of large rotation, displacement and plastic deformation of the fixator components and its joint slippage during experimental testing. The above fact is also a basic requirement for the fixator’s stability in terms of preserving anatomical reduction of bone fragments in the postoperative load conditions.

If we compare the results obtained in relation to the values of stiffness, analyzed Sarafix system configuration (Table 1.), with the results of the stiffness analysis of other external fixators of the same type and similar configuration [2-4, 9], it could be concluded that the Sarafix fixator achieved remarkable results.

From the diagrams (Fig. 5, 8 and 11.) can be seen a good match of results obtained by FEA and experimental analysis. In this way, we can conclude that the solutions obtained by FEA were verified, i.e. the developed CAD/FEM model of the Sarafix fixator was verified. Using the developed FEM model of the Sarafix fixator, it is possible to track 3D displacement of any point of the bone-fixator system and interfragmentary displacements.
within the area of fracture. Using the developed FEM model of the Sarafix fixator, for each case load it is possible to track 3D displacement of any point of the bone-fixator system and interfragmentary displacements within the area of fracture. Due to extreme flexibility of the formed 3D geometrical model, rapid changes were enabled not only to the geometry and position of components and fixator, but also to the materials applied in the external fixation (from stainless steels to radio-transparent composite materials). In this way, conditions for design optimization of the external fixator are created, which would significantly shorten time and reduce development costs of medical devices for external fixation of bones. In addition, the application of such models greatly reduces the volume of conventional preclinical experimental testing of fixators.

The fracture stiffness and Sarafix fixator stiffness have been proven by mechanical research and structural analysis, confirming good clinical results in the treatment of bone fractures. Detailed data of the stiffness of external fixation systems are needed by the orthopaedic surgeon to predict successful healing of a fracture. It is anticipated that this model will provide useful information to surgeons who use Sarafix external fixator for fracture fixation.

It is shown that the CAD/CAE system CATIA can be successfully used in the development of CAD models, FEM analysis and computer simulations of the process from different areas of techniques and medicine. In addition, the application of such models greatly reduces the volume of conventional preclinical experimental testing of fixators.

References

[1] Dozic, S. 2011 „Biomechanical characteristics of unilateral, biplanar fixator Sarafix in the treatment of extensive gunshot-explosive accidental fractures of long bones of the extremities”, Ph.D. Dissertation, Faculty of Medicine Sarajevo, University of Sarajevo, Bosnia and Herzegovina.


