External osteosynthesis with new generation external fixators: focus on innovative construction and simulation computer techniques

J. Deszczynski, W. Choromanski, J. Karpiński

Department of Orthopaedic Surgery, Stępińska 19/25, 00-739 Warsaw, Poland

Abstract

The problem of union of damaged bone fragments is one of the most current topics in modern orthopaedics and the subject of extensive ongoing studies. This article will focus on osteosynthesis using new generation external fixators. New generation implies such solutions that: enable continuous monitoring of the bone union process with the aid of specialized measuring system, allow controlled axial dynamization of the progressively forming union, ensure increased fixator-bone system rigidity due to optimalized spatial arrangement of bone screws and give the possibility of movement in damaged joints or joints localized in the damaged area by the use of mechanical joints reproducing the physiological joint motion pattern. The authors present a novel approach to mathematical modelling of the external fixator-bone mechanical system based on MBS (Multibody Formalism) codes. The concept of modelling and prediction of the process of bone union is also highlighted. Theoretical considerations are exemplified by results of computer simulation and clinical studies.

1. Introduction

External fixation is not a new method of treating bone fractures; on the contrary, its history dates back to the XIX century and is closely linked
with the name of Jean Francois Malgaine. The literature of the subject [2,3,5,6,7] mainly describes concepts of external fixator construction as their properties strongly determine the effectiveness of the therapeutic process. It is not easy to answer how to develop optimal fixator constructions and optimally influence the process of bone union. This article will suggest a solution to the problem and outline a method employing computer-assisted techniques. It appears that computer technology can be applied in the theory and practice of traumatology and orthopaedic surgery, including:

- optimal design of the biomechanical external fixator/bone system;
- monitoring of the therapy, processing of measurement data and developing indications to correct a given therapeutic method (e.g. modification of regulated-dynamization fixator characteristics);
- prognosis of the therapeutic outcome.

2. Description of DYNASTAB-DK fixator construction. Problem definition

Dynastab DK fixators, patented by J. Deszczyński and J. Karpiński, are a family of new generation external fixators dedicated for the treatment of intra- and periarticular fractures of: the knee [fig.16a], ankle [fig.16b], elbow [fig.15] and ulnar as well as fractures of the long bones [fig.13]. The process of setting the bone fragments is helped by the remote control manipulator [fig.14] which is part of the fixator unit. It makes possible the precision location of the fracture by utilizing TV monitors connected to X-Ray machines. The personel is protected from the negative effect of X-Rays. The fixators possess several unique structural properties as they:

1. allow controlled axial interfragmentary dynamization (along the axis of fragment stretching) stimulating bone union and also preventing newly formed delicate tissue structures from crushing - the "mortar effect",
2. assure optimalized spatial arrangement of bone screws which gives the fixator/bone system maximal rigidity, determines good conditions for fracture healing process and enables precise measurement of the interfragmentary interaction forces,
3. include autonomic microprocessor control-measuring system,
4. comprise mechanical joints reproducing physiologic motion within the damaged joints or damaged area which enables early rehabilitation of the diseased extremity.

From the viewpoint of simulation techniques the microprocessor control-measuring system seems most important. The system delivers necessary input data for simulation programs. It is also used to identify the parameters and verify the mathematical models employed. The main concept of the system involves measuring the forces transmitted by the healing bone
and the fixator with different external loads at different stages of healing (fig. 1a). The measure of bone union is defined as follows:

\[ M(t) = \frac{F_1(t)}{F(t)} \times 100\% \]  

(1)

where:

- \( F(t) \) - force load in the extremity,
- \( F_1(t) \) - the part of \( F \) transmitted by the bone,
- \( F(t) = F_1(t) + F_2(t) \),
- \( F_2(t) \) - the part of \( F \) transmitted by the fixator,
- \( t \) - time.

The measure defined above is calculated for various kind of loads (vertical, lateral etc.) Pressure of the bony fragments on the fixator connectors, resulting from loading of the extremity, produces stresses in the fixator's bearing frame, measurable by tensometers located on the frame. The electric signal thus obtained is registered by the board computer of the DYNASTAB DK fixator.

During the process of bone union the first weeks are predominated by osteolysis, i.e. decalcification in the fractured area. In consequence the bone becomes softer and more susceptible to deformations due to external pressure. At the next stage primary union occurs, gradually impregnated with calcium salts. This results in reduced mechanical compliance of the union. At the next stage of bone healing the remodelling of the bone structure takes place during which the union achieves full rigidity and mechanical resistance.

DYNASTAB DK fixator is not only a therapeutic instrument but also a powerful research tool. Its measuring facilities associated with modern software techniques allowed the following scientific objectives to be put forward:

a) Development of optimal spatial arrangement of bone screws ensuring maximal stiffness of the fixator/bone system,
b) Rendering the evaluation of the healing process unbiased,
c) Prediction of the bone union process.

Implementation of the above mentioned research objectives is enabled most importantly by the possibility to detect and register interfragmentary interactions. The DYNASTAB DK board microcomputer possesses a 4 bit CMOS processor, 16 KB Eprom internal memory, 1KB RAM, a tensometric gauge amplifier, a system for producing reference tension, an Eprom memory to memorize configurations, a fixator's joint angle servosystem and a buzzer. Supply voltage: 2.7-6 V, current consumption about 40 \( \mu \)A.

The results memorized by the computer are transmitted to an especially developed steering device which allows to follow them on a built-in display unit. The data are transmissible to a stationary computer. The physician
supervising the therapy can use the options mentioned above with the help of the steering device or a stationary PC-type computer and the patient - by means of the board microcomputer of the DYNASTAB DK fixator.

3. Numeric and experimental methods

The comprehensive analysis of the fixator/bone system uses a numeric system IKARUS 1 (worked out by authors). The general structure of the system is illustrated in figure 2. Pre-processing procedures include:

a) input of geometrical characteristics of the system, types of flexible elements (selection from software library), discretization of bone screws into so called stiffness finite elements [9],
b) input and initial smoothing of measurement data from DYNASTAB DK and update of database of the course of therapy.

Post-processing package mainly consists of module for graphic visualization options.

3.1 Module for the analysis of the physical system - modelling and simulation methods

Most authors employ the method of finite elements to characterize the properties of the bone and bone screws in the field of orthopaedic mechanics. As numeric models are characterized by high level of complexity and difficult identifiability of parameters the extremely simplified beam models are in frequent use. The authors decided to employ MBS codes for the process of modelling [11]. Such approach required discretization of the elements found in the analyzed system and assumption that the system consists of finite number of rigid bodies coupled by flexible elements of diverse characteristics (linear and non-linear). Moreover, the rationale is simplified by the fact that the analyzed system may be taken as quasi-static, therefore most of the inertia forces are insignificant enough to be neglected. The numeric system presented here allows both dynamic and static analysis of linear and non-linear systems. For automatic generation of motion we used Kane equations [8] in the global reference system. They can be simplified as follows:

\[ F^* + F = 0 \]  

(2)

where:

- \( F^* \) is called generalized active forces,
- \( F \) is called generalized inertia forces.

The forces generated by flexible elements are calculated according to the algorithm given by [9]. Optional spatial configuration of flexible elements is assumed. For static analyses of linear systems the numeric program solves a set of linear algebraic equations using the Gaussian method of
elimination. For non-linear cases the program solves a set of differential equations using the Gear's method. Figure 1b illustrates the nominal model used to describe the interaction between the bone screw and the bone. We assumed that the contact phenomena can be characterized by contact stiffness (normal K and tangential KG to the contact area). The stiffness values can be different for any given bone screw. The bending stiffness of finite elements KM was calculated according to [10]. Additionally, the stiffness KZ parameter was introduced in order to determine the fixing stiffness of the bone screw in the fixator frame of DYNASTAB DK [fig.3a]. The method of modelling described above allows the spatial analysis of the motion of the bone fragments and the bone screws at any given load. The spatial configuration of the bone screws is characterized by two angular parameters [fig.3b]:

\[ \beta_1 \] - describing deviation of the axis of bone screws No 1 and 4 from the horizontal plane,

\[ \beta_2 \] - angle - for bone screws No 2 and 3, respectively,

\[ \alpha \] - angle between the bone screws in the horizontal plane. Another parameter of screw situation are provided by the coordinates of their fixing points in the fixator frame.

### 3.2 Experimental studies. Bone union prediction

The experimental studies were performed under both laboratory and clinical conditions. They mainly consisted of generation of the bone union curve according to the measure defined under (2). Prediction of the bone union curve involved determination of its anticipated course based on the following input data:

- data supplied by DYNASTAB DK measuring system: values of the forces transmitted by the fixator at different loads, intensity of rehabilitation exercises, etc.;
- patient age;
- exogenous and endogenous factors;
- type of the fracture (transverse, oblique, comminuted).

Feed forward two-layer "tansig-purelin" neural net was used to predict the course of the process. The first layer contained 50 neurones and the second one - 34. The algorithm of back-propagation was used to train the neural net.

### 4. Example results

#### 4.1 Results of computer simulation of the external fixator-bone physical system

The system highlighted in fig.3b was analyzed for five loads [fig.4]. The values of numerical parameters were derived from table I. Two-dimensional
curves with markers refer to the fixator with spatial configuration of the bone screws \((\alpha = 16^\circ, \beta_1 = 17^\circ, \beta_2 = -17^\circ)\). Interscrew horizontal distance = 50 mm (fig.2). The continuous line refers to the fixator characteristic with linear arrangement of the bone screws \((\alpha = \beta = 0^\circ)\). In this instance successive distances between the screws were 20 mm each. The value of stiffness \(K_G\) (for the load depicted under b) in fig.4) was assumed to be minimal to stress the advantage of spatial configuration of the bone screws. Figures 5-12 show the angular and linear relative displacements of bone fractions in fracture area. Displacement directions refer to load directions. The results confirm that the spatial arrangement of bone screws is beneficial. In the majority of cases the arrangement characterized by \(\beta_1 = 17^\circ\) and \(\beta_2 = 17^\circ\) and \(\alpha = 16^\circ\) proved most advantageous. (The authors call such arrangement asymmetric.) In virtually all cases the results were better than with the fixator coupled to a linear bone screw arrangement. It should be noted that the influence of \(\alpha\) and \(\beta\) angles on the displacements varies with different types of load. Thus \(\beta\) angles importantly affect the reduction in displacements with vertical [fig. 8] and horizontal [fig.9,11] loads. The remaining cases [fig. 10,11,12] illustrate positive influence of \(\alpha\) angle on the construction characteristics.

4.2 Results of clinical studies. An example of a bone union curve
Experimental investigations were performed under laboratory and clinical conditions in the time period from June 1993 to February 1995. Authors have focused mainly on bone union curve determination. The patients group under investigations consisted of seventeen people (8 men and 9 women) - for all of them the treatment methods employing fixators DY-NASTAB DK were used. In the training process of neural network 12 cases were taken into account to create the so called input and target vectors. The values of twelve elements were analyzed in each input vector. They referred to values of \(M(t)\) measures (see relation (2)) received from control-measuring system during the period of treatment and to the others exogenous and endogenous factors. Detailed description of all of them one could find in [1,4]. Output vector from neural network included 34 values of \(M(t)\) measures in the treatment period. Authors employed such neural network to predict bone union curve for fracture of long bone femur. Results presented in figure 17 manifest great efficiency of our method. In spite of rare number of cases used for neural network training the predicted curve 1 is in very good consistence with the real one 2. Curves 1 and 2 refer to patient shown in figure 15. Next the curve 3 (in figure 17) is typical for multifragmentary fractures.
5. Conclusions

Presented was the method of modelling the fixator/bone system using MBS codes. It was employed to optimize spatial configuration of bone screws, called here "asymmetric configuration". It proved to have markedly better characteristics than the linear arrangement of bone screws. Clinical data confirm the results obtained with simulation techniques. The proposed measure and prediction method of bone union is currently at the stage of clinical testing. Therefore it is too early to claim its full utility in clinical practice and research. Nevertheless the simplicity of the method and preliminary results from limited patient group are very promising and encourage further studies. It is our belief that periodic patient follow-up using computer dynamometric scales will allow quantitative evaluation of bone union consolidation and contribute to early detection of its possible disorders. Incorporation of electronics and computer science into the process of fracture healing control is a safe method of imaging the bone union process which will probably enable better knowledge of the mutual relations between interfragmentary forces and the consolidating bone union. Such knowledge may prove an important factor in the understanding the impact of physical phenomena on the biology of the bone.

Acknowledgement

This research is supported by Polish State Committee for Scientific Research (KBN) - grant no. 3 P401 033 06

References:


\[ F = F_1 + F_2 \]

\[ \text{Figure 1: a) Illustration of bone union measure b) Implant-bone interaction} \]
Figure 2: Simplified scheme of program IKARUS

Figure 3: a) Nominal model of the bone screw b) Schematic representation of spatial configuration of bone screws.
Figure 4: The analyzed loads: a) Vertical force $F_z$, b) Lateral force $F_y$, c) Bending moment $M_z$, d) Torsional moment $M_x$, e) Torsion force $F_x$.

Figure 5: Torsion $F_x$ and lateral $F_y$ loads

Figure 6: Vertical $F_z$ and torsional $M_x$ loads
Figure 7: Bending $M_z$ load

Figure 8: Vertical $F_y$ load

Figure 9: Lateral $F_y$ load
Figure 10: Torsion $F_z$ load

Figure 11: Torsional $M_z$ load

Figure 12: Bending $M_z$ load
Figure 13: a) Patient during examination with DYNASTAB-DK measuring-control system, b) X-Ray taken from the patient with femur fracture from fig.13a

Figure 14: DYNASTAB DUO remote control during fracture reposition
Figure 15: a,b) DYNASTAB DK fixator in functional fracture treatment, c,d) X-Ray of fractures from fig.15a and 15b
Figure 16: DYNASTAB DK fixator for: a) articular knee fractures treatment, b) articular ankle fractures treatment.

Figure 17: Curve showing progress of fracture healing - results obtained from neural network software.
<table>
<thead>
<tr>
<th>Symbol</th>
<th>Value</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>$K_z$</td>
<td>$1E6 \text{ N/m}$</td>
<td>Implant fixing stiffness in the fixator frame</td>
</tr>
<tr>
<td>$KG$</td>
<td>$1E6 \text{ N/m}$</td>
<td>Tangential contact stiffness</td>
</tr>
<tr>
<td>$K$</td>
<td>$1E6 \text{ N/m}$</td>
<td>Normal contact stiffness</td>
</tr>
<tr>
<td>$E$</td>
<td>$2.0E11 \text{ N/m}$</td>
<td>Young modulus for the bone screw material</td>
</tr>
<tr>
<td>$D$</td>
<td>$3.0 \text{ m}$</td>
<td>Averaged bone diameter</td>
</tr>
<tr>
<td>$l$</td>
<td>$0.09 \text{ m}$</td>
<td>Length of the bone screw measured from fixing point</td>
</tr>
<tr>
<td>$\alpha$</td>
<td>$0 - 16 \text{ deg}$</td>
<td>Analysed $\alpha$ angle range</td>
</tr>
<tr>
<td>$\beta$</td>
<td>$\pm 17 \text{ deg}$</td>
<td>Analysed $\beta$ angle range</td>
</tr>
<tr>
<td>$F_z$</td>
<td>$500 \text{ N}$</td>
<td>Vertical load (for three dimensional graphs)</td>
</tr>
<tr>
<td></td>
<td>$10 - 500 \text{ N}$</td>
<td>Range of loads (for two dimensional graphs)</td>
</tr>
<tr>
<td>$F_t$</td>
<td>$300 \text{ N}$</td>
<td>Torsional load (for three dimensional graphs)</td>
</tr>
<tr>
<td></td>
<td>$10 - 300 \text{ N}$</td>
<td>Range of loads (for two dimensional graphs)</td>
</tr>
<tr>
<td>$F_y$</td>
<td>$300 \text{ N}$</td>
<td>Lateral load (for three dimensional graphs)</td>
</tr>
<tr>
<td></td>
<td>$10 - 300 \text{ N}$</td>
<td>Range of loads (for two dimensional graphs)</td>
</tr>
<tr>
<td>$M_z$</td>
<td>$40 \text{ Nm}$</td>
<td>Bending moment (for three dimensional graphs)</td>
</tr>
<tr>
<td></td>
<td>$1 - 50 \text{ Nm}$</td>
<td>Moment range (for two dimensional graphs)</td>
</tr>
<tr>
<td>$M_t$</td>
<td>$40 \text{ Nm}$</td>
<td>Torsion moment (for three dimensional graphs)</td>
</tr>
<tr>
<td></td>
<td>$1 - 50 \text{ Nm}$</td>
<td>Moment range (for two dimensional graphs)</td>
</tr>
<tr>
<td>$\beta_2$</td>
<td>$0 \text{ deg}$</td>
<td>Value for three dimensional graphs with $\beta_1$ and $\alpha$ as variables</td>
</tr>
<tr>
<td>$\alpha$</td>
<td>$16 \text{ deg}$</td>
<td>Value for three dimensional graphs with $\beta_1(2)$ as variables</td>
</tr>
</tbody>
</table>