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EXTERNAL MECHATRONIC ORTHOPAEDIC FIXATORS – MODELLING, COMPUTER SIMULATIONS AND CLINICAL EXPERIENCES

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This paper presents some selected issues of modelling and simulation of physical performance of the External Unilateral Mechatronic Orthopaedic Fixator-Bone (EUMOFB) system intended for treatment of shafts of long bones. The model structure includes both a nominal model of the system and a mathematical model design technique. The research inquiry has been limited to discrete models. While considering a clinic postulate of functional treatment of fractures, it has been introduced a stability concept in technical terms. A monitoring method for fracture healing process where both measures of bone union and heuristic techniques depending on artificial feedforward neural networks have been employed is presented too. The test readings were used to build fixators of a new generation. Some results obtained in environment of a clinic are also presented in this paper.

Key words: modelling, computer simulations, orthopaedic fixators

1. Introduction

External orthopaedic fixators are employed for treatment of bone fractures (those compound ones, in particular comminuted fractures, in which probability of complications is great if a conventional treatment is employed, also the fractures for which other treatment methods have produced negative effects – pseudarthrosis or lack of bone union). A fixator consists of two principal components; i.e., bearing frame that is fixed externally to the patient's body and bone screws that secure the bearing frame to the bone fragments. A variety of bone screw bearing frame structures defines particular designs of fixators whose examples are shown in Fig.1A. For instance, see De Bastini and Aldegheri (1986), Będziński (1997), Canadell and Forriol (1993), Goodship et al. (1993), Lane (1987) for a survey of various fixator structures. From a clinical view point (where easy installation and low risk of damage to nerves etc. are taken into account), the so-called unilateral fixators become more and more popular (Fig.1A(a)). The latest designs enable the functional treatment postulate to be observed (De Bastini and Aldegheri, 1986; Canadell and Forriol, 1993; Deszczyński and Karpiński, 1992; Jasińska-Choromańska et al., 1996). This postulate consists in generation of micro-movements in a direction that is stricly determined, i.e. the axial one with respect to the bone, the best rigidity of the system being maintained for the remaining directions in these structures employed for healing the fractures of long bones.

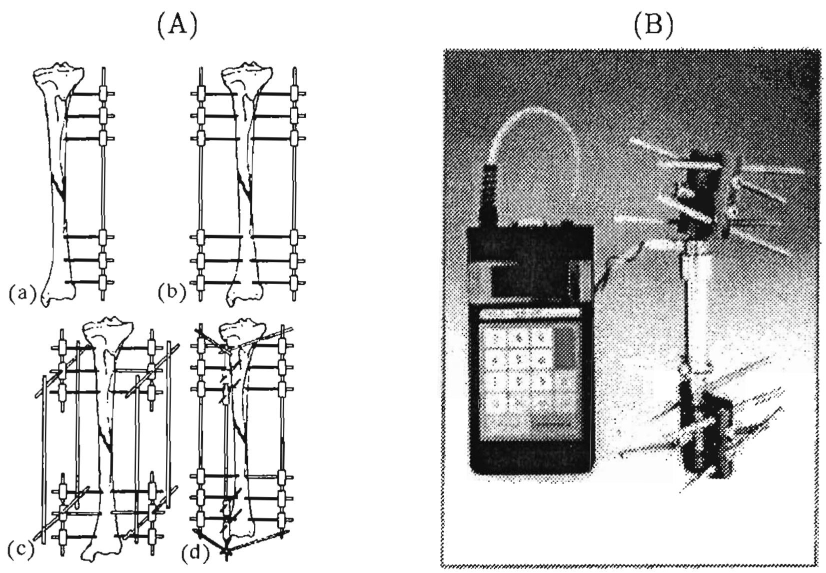

Fig. 1. (A) – Spatial configurations of external fixators: (a) unilateral, (b) bilateral, (c) quadrilateral, (d) triangular; (B) – DYNASTAB DK ultra-modern external fixator equipped with a dynamisation chamber and a computer system for monitoring the bone-union processes

These micro-movements stimulate the osteogenic processes. Values of micro-movements should be under adaptive control in the treatment process. The capability of generation of micro-movements during the treatment process has been called Treatment Made Dynamic, while the structure intended to achieve this goal – Dynamisation Chamber. The Type DYNASTAB DK Ultra-modern External Fixator (Deszczyński and Karpiński, 1992) designed for healing long-bone shaft fractures is shown in Fig.1B. Computer-aided simulation techniques have been used for sub-optimisation of the design of this fixator. These techniques are discussed elsewhere in this paper.

2. Nominal model

The process of building the mathematical model of EUMOFB has required a conceptual model (i.e. an ideal one) to be developed, which is to be called a nominal model elsewhere in this paper, and to describe its properties by using this basis. The mathematical description of nominal model understood as a set of relationships and formulae describing the phenomena occurring in the machine, plant or process is called a mathematical model. The nominal model is a system made of physical relationships selected on the basis of the identified structure of the machine, plant or process, the properties of its individual components, and external actions (e.g. forced inputs). A real system can be analysed in accordance with various criteria and phenomena considered; therefore, a number of nominal models can be proposed for the system under investigation with respect to particular objectives. And, in turn, a number of mathematical models depending, for instance, on the assumed simplifications and used formalisms of the analytical mechanics can be developed for each nominal model. In our case, the criterion that is the most significant with respect to nominal-model analysis and selection is to take those features of the real system, whose effect upon the phenomenon under investigation is crucial. Determination of the effect of spatial configuration of the bone screws on the stability of bone fragments and analysis of dynamisation chamber properties in terms of the postulate of functional treatment of fractures have been among others one of the major testing objectives within the scope of mechanical phenomena; the term of stabilisation means that a constant relative position of bone fragments is kept within specified accuracy limits in situations where admissible mechanical loads are applied to the system. It is accepted that a structure featuring highest rigidity and if the micro-movement values are adjustable only by the dynamisation chamber is the best structure. A nominal model of the system is shown in Fig.2.

The spatial configuration of the bone screws is identified by the angles $alpha$, β_1 and β_2. For the purpose of analysis of screws, the ideal rigidity of the fixator frame has been assumed. This is justified to a great extent in practice. The frame in the case of the family of the Type DYNASTAB Fixators is a pipe having an outside diameter of 20 mm, and can be accepted

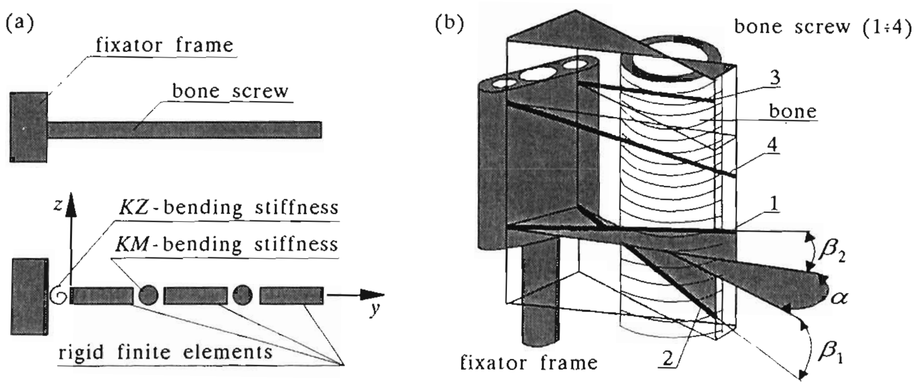

Fig. 2. (a) Nominal model of the bone screws, (b) nominal model of the external fixator-bone system (fixator frame assumed as ideal rigid)

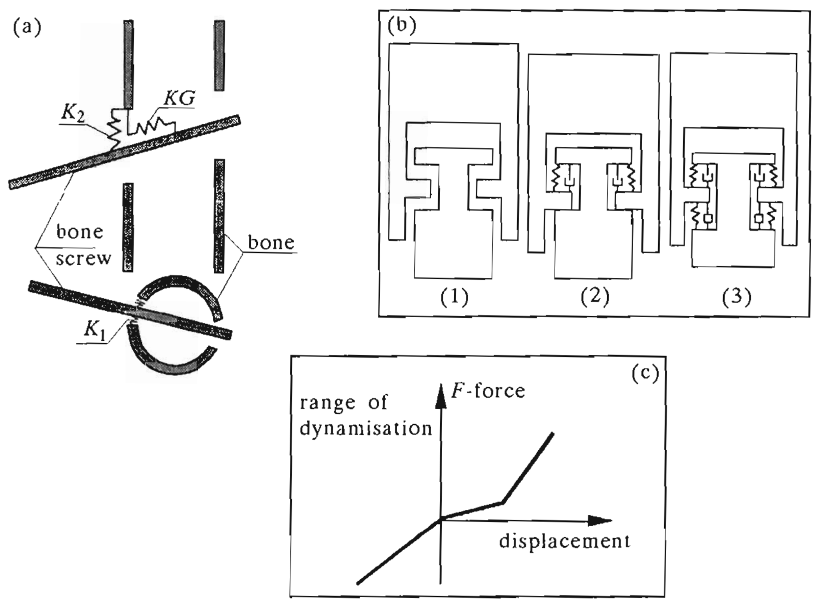

Fig. 3. (a) Nominal model of the implant (bone screw) affecting the bone; (b) model of dynamisation chamber: 1) two directional with clearence, 2) unidirectional with flexible element, 3) two directional with flexible element; (c) elastic characteristics of the chamber

as a component that is non-deformable in the case when the real loads act under clinical conditions. The method of rigid finite elements (Kruszewski et al., 1975), which had been modified by the author to take the non-linear characteristics of flexible elements, was employed for modelling the bone screws (however in the present paper only linear and bilinear characteristics have been analysed). The method depend on dividing the real systems, including the continuous ones, into non-deformable solids that are called the rigid finite elements. A number of advantages feature the method of rigid finite elements. Possibility of the use of a variety techniques of mechanics and simulation developed for discrete systems is one of them. The accepted nominal model of the bone screws is shown in Fig.3a. It has been assumed that the screw material and the bone fragment tissue affect each to the other elastically in the normal and tangential directions with respect to the contact area in the nominal model of a screw-bone system (cf Fig.2a). A nominal model of the dynamisation chamber is shown in Fig.3b, while the values of the characteristics are listed in Table 1.

Table 1. Values of the model parameters[1]

Desig.	Value	Definition
KZ	$1 \cdot 10^5$ Nm/rad	stiffness of screw mounting in the bearing frame
K_1	Calcu-lated	bending rigidity between the finite elements modelling the bone screw (see Fig.3)
KG	$1 \cdot 10^6$ N/m	tangential rigidity in a contact region (see Fig.3)
K_2	$1 \cdot 10^5$ N/m	rigidity in a contact region in normal direction (see Fig.3)
E	$2.0 \cdot 10^{11}$ N/m^2	Young modulus of the materials, which the frame and screw bones are made of
d	0.006 m	root diameter of the bone screw
R	0.05	distance between the point of mounting the screw in the fixator frame and the nearest point of contact between the screw and bone
D	0.03 m	averaged diameter of bone

[1]Values of parameters from unpublished works carried out by author (grant of Dean of Faculty) – rapport entitled "Analysis of mechanical and clinical properties of the bone- external fixator system" available in Institute of Micromechanics and Photonics, Faculty of Mechatronics, Warsaw University of Technology, 1996

0.5α	17 deg	value of the angle 0.5α in spatial charts with various values of the angles β_1 and β_2
	$0 \div 20$ deg	range of variations of the angle 0.5α
β	± 17 deg	range of variations of the angles β_1 and β_2
β_2	0 deg	value of the angle β_2 in spatial charts with various values of the angles β_1 and α
F_x, F_y	300 N	horizontal loads in spatial charts
	$0 \div 300$ N	range of variations of the load in plane charts (see Fig.7)
F_z	500 N	vertical load in spatial charts
	$0 \div 500$ N	range of variations of the load in plane charts (see Fig.7)
$M_x, M_y,$	10 Nm	loading torques in spatial charts
M_z	$0 \div 20$ Nm	range of variations of the load in plane charts (see Fig.7)
M	50 kg	reduced mass of the patient body above the fracture, analysed in dynamic studies
J_x, J_y, J_z	$20, 20, 7$ kg m^2	reduced moment of inertia of the patient's body above the fracture
Kr	$8 \cdot 10^6$ N/m	compression rigidity of the bearing frame of the fixator
Kd	$0 \vee 10^4 \vee \infty$ N/m	axial rigidity of the dynamisation chamber
L	$0 \vee 0.002$ m	clearance in the dynamisation chamber

3. Methods of modelling and simulations

3.1. Equations of motion

First, let us assume that the general form of the equations of motion for the EUMOFB system can be expressed as follows

$$\mathsf{A}(\boldsymbol{p})\ddot{q} + \boldsymbol{F}(\boldsymbol{p}, \dot{q}, \boldsymbol{q}) = \boldsymbol{Q}_w \tag{3.1}$$

where

$\mathsf{A}(\boldsymbol{p})$ – matrix of inertia

$\boldsymbol{p}$ – vector of parameters

$\boldsymbol{F}$ – vector of forces produced by flexible parameters
$\boldsymbol{Q}_w$ – forcing inputs affecting the system.

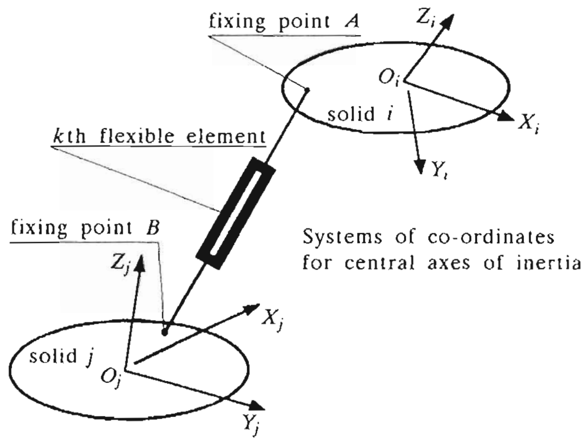

Fig. 4. System of two rigid solids coupled by a flexible element

Let us assume that the elements of EUMOFB system vibrate at a small amplitude around the static balance position. And for the purposes of our simulations we are interested just in this static balance position and the values of micro-displacements around this position. As a rough approximation, we can accept this system as a system without geometrical constraints. As previously, let us focus our attention on determination of the forces associated with flexible elements (associated with rigidity features listed elsewhere in Section 2). The investigation is conducted for a single element but this does not limit the general problem solving at all. Let us assume that a kth flexible element couples both solids (which are seen as non-deformable) with the indices i and j, respectively (see in Fig.4). Let us designate the point at which the kth element is fixed to the ith solid with symbol A and that to the jth solid with symbol B. The axis which couples the points A and B is to be called a kth flexible element axis. The solids under investigation can be elements of a greater mechanical system which are identified generally (without mechanical constraints taken into consideration) by the vector of state

$$\boldsymbol{x} = [\boldsymbol{q}, \dot{\boldsymbol{q}}]^{\top} \tag{3.2}$$

where
$\boldsymbol{q}$ – vector of displacements $(1 \times 6N)$
$\dot{\boldsymbol{q}}$ – vector of velocity $(1 \times 6N)$
N – number of solids in the system under investigation.

The motion of ith solid is identified by means of

$\boldsymbol{Q}_{pl}^{(i)}$ – three-element vector of translational displacements (along the axes $0X$, $0Y$, $0Z$, respectively, as shown in Fig.4)

$\boldsymbol{Q}_{pr}^{(i)}$ – three-element vector of rotational displacements (around the axes $0X$, $0Y$, $0Z$, respectively, as shown in Fig.4)

$\boldsymbol{Q}_{vl}^{(i)}$ – three-element vector of translational velocities

$\boldsymbol{Q}_{vr}^{(i)}$ – three-element vector of rotational velocities.

The relation between individual lth component of these vectors and those of the $\boldsymbol{x}$ state vector is as follows

$$\begin{aligned} (Q_{pl}^{(i)})_l &= x_{6(i-1)+l} \\ (Q_{pr}^{(i)})_l &= x_{6(i-1)+l+3} && l = 1,2,3 \\ (Q_{vl}^{(i)})_l &= x_{6(i-1)+l+6N} && i = 1,2,...,N \\ (Q_{vr}^{(i)})_l &= x_{6(i-1)+l+3+6N} \end{aligned} \tag{3.3}$$

where index l refers to the co-ordinate system directions x, y and z, respectively.

To determine the forces generated by the flexible elements (these elements are passive ones such as springs, dampers, etc. for the purposes of conventional analysis) it is necessary to determine their strains and velocities of strains (where the term "strain" is understood as a difference between translational and rotational displacements of the points where an element is fixed to the ith and jth solids. To this end, it is necessary to calculate the fixing point displacements and the velocities of these displacements (where the displacements and velocities of points A and B in Fig.4 are considered) at the first stage.

Let us perform an analysis for the point A.

Let the co-ordinates of this point in the system associated with the ith solid be identified by means of a three-element vector $\boldsymbol{Z}_{am}^{(ik)}$.

Then the vector $\boldsymbol{Z}_{pl}^{(ik)}$ of the translational displacements of the fixing point in the system associated with the ith solid is defined by the following relation

$$\boldsymbol{Z}_{pl}^{(ik)} = \boldsymbol{Q}_{pl}^{(i)} + \mathsf{M}^{(ik)}\boldsymbol{Q}_{pr}^{(i)} \tag{3.4}$$

where

$\mathsf{M}^{(ik)}$ – skew-symmetric matrix

$$\mathsf{M}^{(ik)} = \begin{bmatrix} 0 & (Z_{am}^{(ik)})_3 & -(Z_{am}^{(ik)})_2 \\ -(Z_{am}^{(ik)})_3 & 0 & (Z_{am}^{(ik)})_1 \\ (Z_{am}^{(ik)})_2 & -(Z_{am}^{(ik)})_1 & 0 \end{bmatrix}$$

and

$(Z_{am}^{(ik)})_l$ – co-ordinates of the fixing point of kth flexible element associated with the ith solid, $l = 1, 2, 3$.

Similarly, the vector of velocities of the fixing point is defined by the following relation

$$\boldsymbol{Z}_{vl}^{(ik)} = \boldsymbol{Q}_{vl}^{(i)} + \mathsf{M}^{(ik)}\boldsymbol{Q}_{vr}^{(i)} \tag{3.5}$$

Let the components of three-element vector $\boldsymbol{K}_{cos}^{(ik)}$ define the cosines of angles between the axis of the flexible element under investigation and successive axes of the system of co-ordinates associated with the ith solid. Then the value of translational displacement the of fixing point of the kth flexible element the solid A when measured along the axis of the element is defined by the following relation

$$S_{pl}^{(ik)} = \boldsymbol{Z}_{pl}^{(ik)\top}\boldsymbol{K}_{cos}^{(ik)} = \sum_{l=1}^{3}(Z_{pl}^{(ik)})_l(K_{cos}^{(ik)})_l \tag{3.6}$$

In a similar way we can define:

— velocities

$$S_{vl}^{(ik)} = \boldsymbol{Z}_{vl}^{(ik)\top}\boldsymbol{K}_{cos}^{(ik)} = \sum_{l=1}^{3}(Z_{vl}^{(ik)})_l(K_{cos}^{(ik)})_l \tag{3.7}$$

— rotations

$$S_{pr}^{(ik)} = \boldsymbol{Q}_{pr}^{(ik)\top}\boldsymbol{K}_{cos}^{(ik)} = \sum_{l=1}^{3}(Q_{pr}^{(ik)})_l(K_{cos}^{(ik)})_l \tag{3.8}$$

— angular velocities

$$S_{vr}^{(ik)} = \boldsymbol{Q}_{vr}^{(ik)\top}\boldsymbol{K}_{cos}^{(ik)} = \sum_{l=1}^{3}(Q_{vr}^{(ik)})_l(K_{cos}^{(ik)})_l \tag{3.9}$$

Similar displacements and velocities can be defined for the other coupled solid (jth). The translational and rotational strains as well as the velocities of the strains can be defined by the following relations

$$\begin{aligned}
D_{S_{pl}}^{(jik)} &= S_{pl}^{(jk)} - S_{pl}^{(ik)}\\
D_{S_{pr}}^{(jik)} &= S_{pr}^{(jk)} - S_{pr}^{(ik)}\\
D_{S_{vl}}^{(jik)} &= S_{vl}^{(jk)} - S_{vl}^{(ik)}\\
D_{S_{vr}}^{(jik)} &= S_{vr}^{(jk)} - S_{vr}^{(ik)}
\end{aligned} \tag{3.10}$$

It can be seen that all the components $S_{pl}^{(jk)}$, $S_{pr}^{(jk)}$, $S_{vl}^{(jk)}$, $S_{vr}^{(jk)}$ are equal to zero in the cases when the considered element is joined at one its end to a support.

The values of the force and torque vectors $\boldsymbol{R}^{(k)}$, $\boldsymbol{F}^{(k)}$ that are generated by strains and strain velocities in the kth flexible elements are defined by the following relations

$$
\begin{aligned}
R^{(sk)} &= f_a(D_{S_{pl}}^{(jik)}, D_{S_{pr}}^{(jik)}, D_{S_{vl}}^{(jik)}, D_{S_{vr}}^{(jik)}) \\
F^{(sk)} &= f_b(D_{S_{pl}}^{(jik)}, D_{S_{pr}}^{(jik)}, D_{S_{vl}}^{(jik)}, D_{S_{vr}}^{(jik)})
\end{aligned}
\tag{3.11}
$$

where $s = i,j$ and f_a, f_b – stand for flexible element characteristics (for example a linear characteristic).

Now, let us define the components of forces and torque generated by a flexible element along a system that is associated with main central axes of inertia of the solids.

The components of vector $\boldsymbol{F}^{(ik)}$ (value of which is representedd by Eq (3.11)) in the system associated with the ith and jth solids are defined by the following relations

$$
(F^{(sk)})_l = F^{(sk)}(K_{cos}^{(sk)})_l \qquad \begin{array}{l} s = i,j \\ l = 1,2,3 \end{array}
\tag{3.12}
$$

It can be seen that in situations where the direction of the force does pass through the centre of the solid mass, the force generates the torque that is equal to the vector product of the force and the radius-vector determining the point where the force is applied. And finally the torque vector is defined by the following relations

$$
\mathbf{M}^{(sk)} = \mathbf{M}(k) + \boldsymbol{Z}_{am}^{(sk)}\boldsymbol{F}^{(sk)} \qquad \begin{array}{l} s = i,j \\ l = 1,2,3 \end{array}
\tag{3.13}
$$

This procedure is repeated for all flexible elements. The methods presented above have been employed for automatic generation of the equations of motions.

3.2. Analysis of technical stability

The postulate of functional treatment of long-bones depends on such shaping of the EUMOFB system that it is capable of performing micro-movements of the bone fragments in a direction and a range that are strictly determined (see Section 1) and this stimulates the osteogenic processes. From the author's standpoint, this postulate is strictly associated with the stability in terms of technical stability. In the literature, the latter term has been associated with the names of such R&D people as Moisieiev and Chetaiev and was discussed the two excellent monographs by Skalmierski and Tylikowski (1973) and Bogusz (1972). The latter monograph was employed for investigating the technical stability in this paper. To present the definition of the technical stability, let us consider the differential equation expressed by Eq (3.1) with the initial condition $\boldsymbol{q}(t_0) = \boldsymbol{q}_0$ and $\dot{\boldsymbol{q}}(t_0) = \dot{\boldsymbol{q}}_0$. During the process of simulation the vector $\boldsymbol{Q}_w$ can be regarded as a deterministic or stochastic force input.

If each solution of Eqs (3.1) with an initial condition belonging to an area ω belongs to an area Ω, the set of equations (3.1) is stable technically with respect to areas the $\omega \subset \Omega$ as well as the input $\boldsymbol{Q}_w$. The area Ω determines the area of admissible values of the generalized co-ordinates, while the $\boldsymbol{Q}_w$ determines the admissible loads of the EUMOFB system under clinical conditions. By taking the specific features of the investigated system into consideration, the technical stability can be supplemented with a postulate of solutions located at or "near" boundary of the area Ω in the cases when the "highest" loads occur. If this condition is met, the bone union is then not over-rigid, and the micro-movements stimulate the osteogenic processes. In this way, we are capable of optimising the values of selected parameters at the designing stage to provide proper dynamics of the bone uniting process. To this end let us define a quality coefficient in the following form

$$Q = \int\limits_0^T \Big[\sum_{i=1}^{N} q_i^2 + (P_{allow} - |q_k - q_l|)^2\Big]\, dt \tag{3.14}$$

where

P_{allow}	–	admissible relative displacement for a particular force input Q_w
Q_l, q_k	–	values of displacements for a selected direction (axial with respect to a long bone)
$\lvert q_k - q_l\rvert$	–	absolute relative displacement between bone fragments in the selected direction
T	–	investigation time (time of simulation of the EUMOFB system).

The optimization problem can be formulated as follows:

- Let us find the optimal values of given parameters (in Eqs (3.1)) to minimise the quality coefficient (3.14).

Any selected parameters of the model can be the variables (for instance, the diameters of bone screws) included in the optimisation process. Various optimisation techniques can be employed in the optimisation process. In this paper, the Fletcher Powell's variable metric method was used (Jasińska-Choromańska et al., 1996).

3.3. Basic idea of bone uniting process analysis

As it was noted in the Introduction, the assessment of bone uniting process with the use of conventional methods of clinical practice, i.e. employing x-ray pictures and manual examination is more or less subjective and liable to fail. Making this assessment objective and allocating some measure to it, to identify the advance in the bone uniting process quantitatively, is a significant issue of computer-aided investigation. The idea of the bone union measure depends on measuring the total load acting onto the healed limb as well as that acting on the bearing frame of the fixator F_2 and that acting on the bone being united F_1. With the values mentioned above being known, one can introduce a measure of the bone union

$$M = \frac{F_2}{F} = \frac{F_2}{F_1 + F_2} \tag{3.15}$$

It can be seen that the measure is function of time. And its progress in time defines the so-called bone union response. The concept of the invention depends on depends on monitoring of the bone union response and developing such methods on this basis that a standard bone union response and healing process diagnostics (deviations from the bone union response) become available. Analysis of the bone union responses for various cases will enable the physicians to investigate the effects of various factors on the bone uniting processes (thus, it becomes an important research and development tool). The load acting upon the fixator-bone system due to an axial force is shown in Fig.5.

It should be kept in mind, however, that the mechanical properties of a bone being united may be strongly non-linear and therefore the bone union measures can appear different for both various values of a particular load as well as for various loads (due to forces acting in various directions and torque). Therefore, it seems that there are grounds for introducing not a single measure but rather an n-element vector of bone union measures and the vector of standard responses

$$M_i = \frac{F_{2i}}{F_i} = \frac{F_{2i}}{F_{1i} + F_{2i}} \qquad i = 1, ..., n \tag{3.16}$$

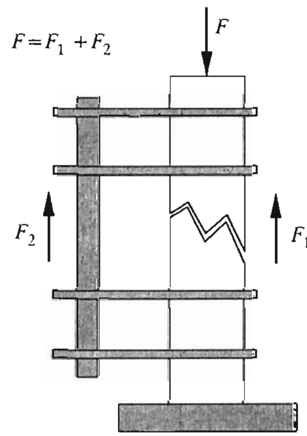

Fig. 5. Concept of the bone union measure

Positive result of treatment leads to a situation where all the bone union measures go to zero, i.e. that the total load is carried by the limb. Selection of a method for analysing the bone union responses provided by the DYNASTAB DK Fixator test system becomes an important issue. Multiplicity of cases (age of seek people, type of fracture, other diseases, intensity of rehabilitation exercises, used medicines, etc.) result in difficult selection of computer selection techniques. The techniques using the artificial intelligence approach, namely, artificial non-linear and multi-layer neurone networks. have been finally choosen. A proposal is to distinquish 12 fracture types. A particular neural structure was tailored for each fracture type and every gender and every addiction (details are presented by Jasińska-Choromańska et al., 1997).

Although each structure has been developed on the basis of feedforward neural network (Tadusiewicz, 1993) owing to introduction of feedback approach its real structure has been rather that of a recurrent neurone network. The architecture of neural predictor intended to predict the bone union responses is shown in Fig.6.

The existing treatment condition (T_i State) is identified by seven bone union responses determined for various loads in compliance with the measures obtained from the DYNASTAB DK system or from the previous calculation step in the computer-aided simulation process. The parameters designated with parameteres 1 in Fig.6 indicate the values of selected analyses: levels of Ca and P in serum, base phosphotase, intensity of rehabilitation (as measured with the use of a measuring fixator system), time passing since the fixator has been installed. And the parameters designated with parameters 2

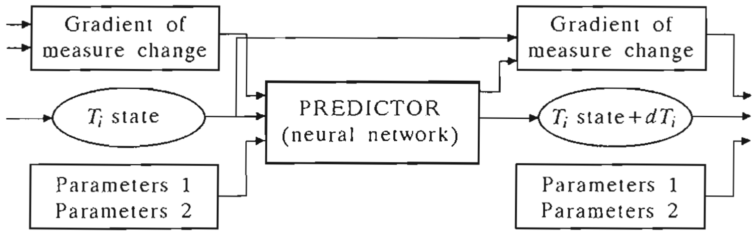

Fig. 6. Architecture of a system intended to predict the bone union responses

in Fig.6 indicate the age of the patient, bone density, dT_i prediction interval, operation technique employed (e.g. stabilisation with initial compression). A structure of non-linear three-layer neural network has been employed for investigation where the results of clinical examinations and algorithm of error backpropagation have been used for the training method. The network has taken a form of three-layer neurone network mentioned above including 617 neurones in total (the first layer of 40 neurones to provide transformation by means of hyperbolic tangent function, the second layer of 570 neurones to provide transformation by means of logistic function, and the third layer of 7 neurones as this is the number of outputs from the system associated with seven bone-union measures, which is intended to provide transformation by means of logistic function too). To train the network, the algorithm of error backpropagation with modified torque has been used. The following load types have been employed to determine seven bone-union measures in the case of thigh bone fracture:

M_1 – contraction of thigh muscles for dorsal decubitus
M_2 – active lifting of the limb to a position at an angle 30 degrees
M_3 – limb is lifted passively and the muscles are relaxed
M_4 – contraction of muscles when the limb is in its vertical position
M_5 – lateral recumbency on healthy side when the limb is abducted by an angle of 20 degrees
M_6 – 100 N load in vertical position
M_7 – 200 N load in vertical position.

for prediction of evolution of bone-union measures on the basis of data provided in tests made at initial stage of treatment.

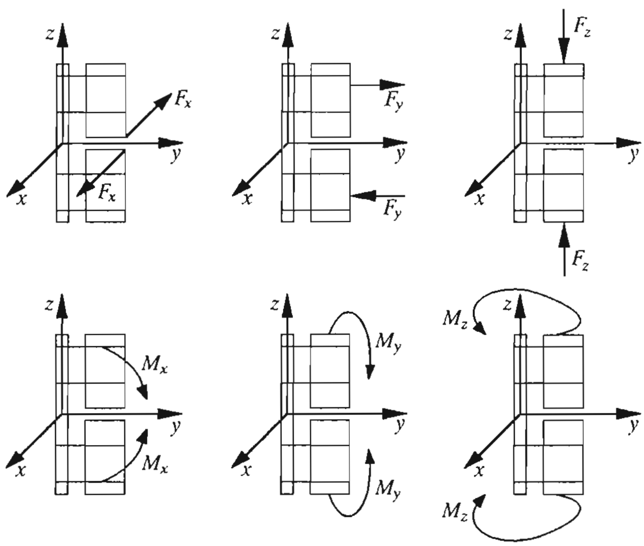

Fig. 7. Analysed types of loads in the external fixator-bone system

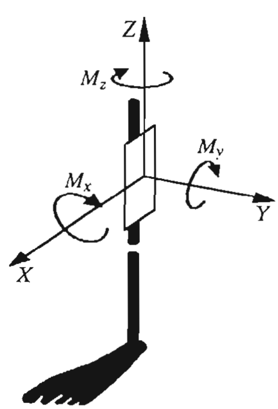

Fig. 8. Analysed types of loads in the external fixator-bone system

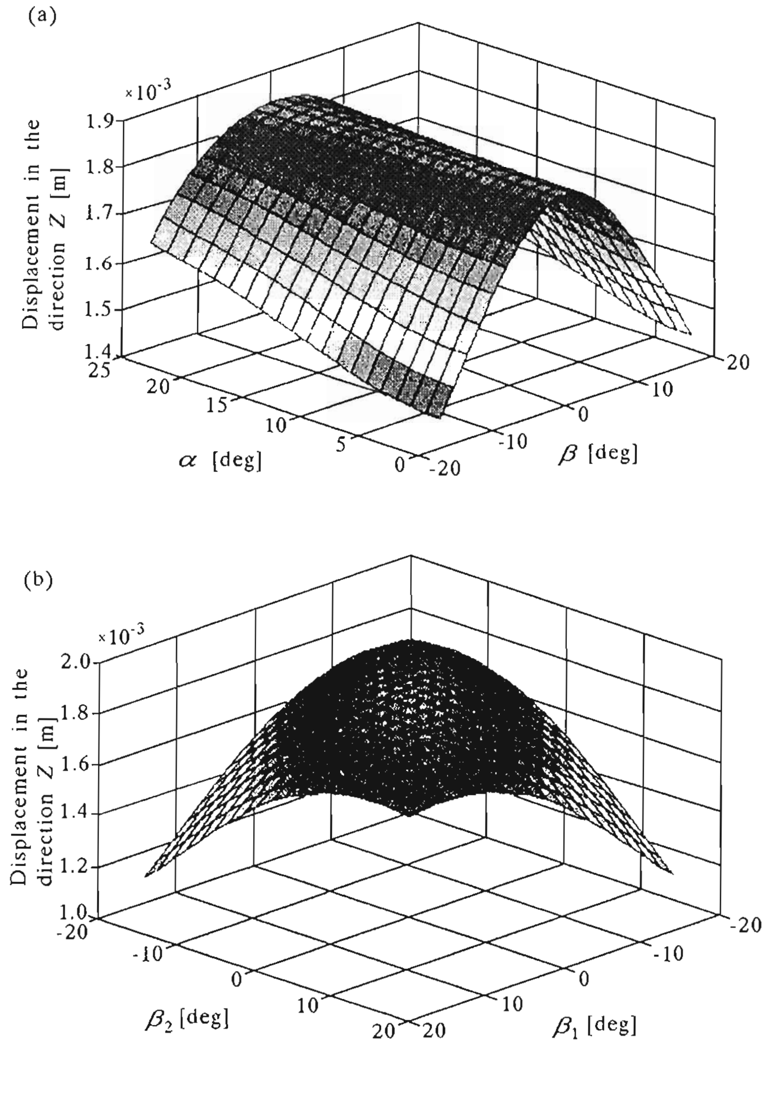

Fig. 9. Results of computer simulations: (a) F_z – load (relative translational displacement in the direction Z of bone fraction versus values of angles β and α), (b) F_z – load (relative translational displacement in the direction Z of bone fraction versus values of angles β_1 and β_2)

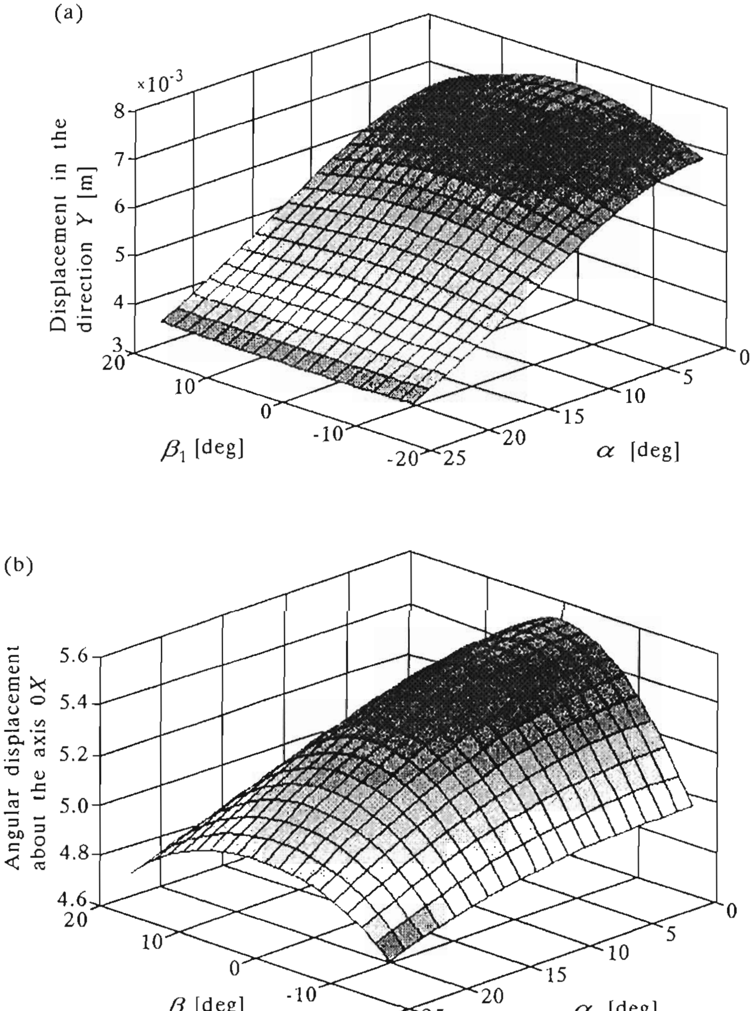

Fig. 10. Results of computer simulations: (a) F_y – load (relative translational displacement in the direction Y of bone fraction versus values of angles β_1 and α), (b) M_x – load (relative rotation displacement about the axis $0X$ of bone fraction versus values of angles β and α)

4. Sample simulation results

4.1. Static tests – fixing rigidity of bone screws

The tests were performed for a nominal model similar to that discussed in Section 2 and a model developed in compliance with the method presented in Section 3 was used. The types of local are shown in Fig.7 (for orientation of the system of co-ordinates see Fig.8). The exemplary simulation results are shown in Fig.9 and Fig.10. A positive effect of "spatial tent" configuration of bone screws upon the system rigidity with respect to the linear configuration (where α and β are equal to zero) is evident. The following specifications have been accepted for an optimal system: $\alpha = 20$ degrees, $\beta_1 = 17$ degrees, $\beta_2 = -17$ degrees. The limitations imposed on the values of angles have resulted from some clinical circumstances and a significant rise in the tangential stresses in the contact areas occurring with the rise in the values of angles β.

4.2. Dynamic tests – technical stability of the system

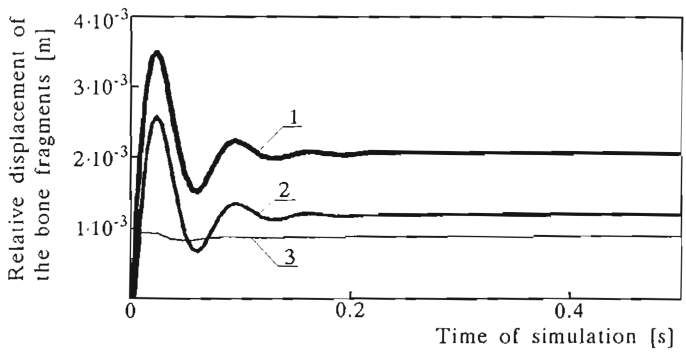

Fig. 11. Simulation results: the system before the optimising procedure under the static vertical force $F_z = 500\,\mathrm{N}$; 1 – total displacement, 2 – component of the bone-screw strain, 3 – component of the dynamisation chamber displacement

An analysis of a system employing a dynamisation chamber was performed. A bone-screw core diameter of 4.5 mm has been assumed (as a starting value). The damping coefficient of soft tissues in the system has been assumed as $c = 2000\,\mathrm{kg/s}$. It is difficult to identify the latter parameter and substantial scatter can occur in practice. Simulation examples are presented in Fig.11, Fig.12 and Fig.13. In the first test the static vertical force $F_z = 500\,\mathrm{N}$ was applied. The relative displacement of the bone fragments in the vertical direction is a sum

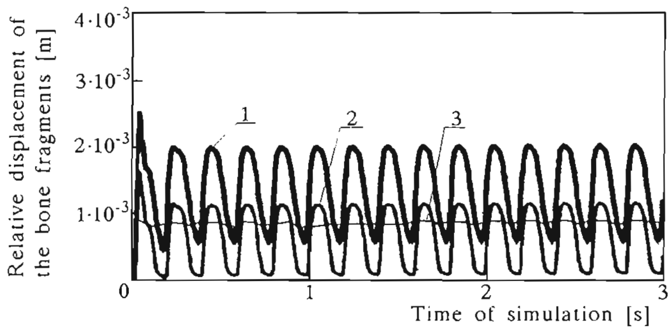

Fig. 12. Simulation results: the system before the optimising procedure under a variable force; 1 – total displacement, 2 – component of the bone-screw strain, 3 – component of the dynamisation chamber displacement

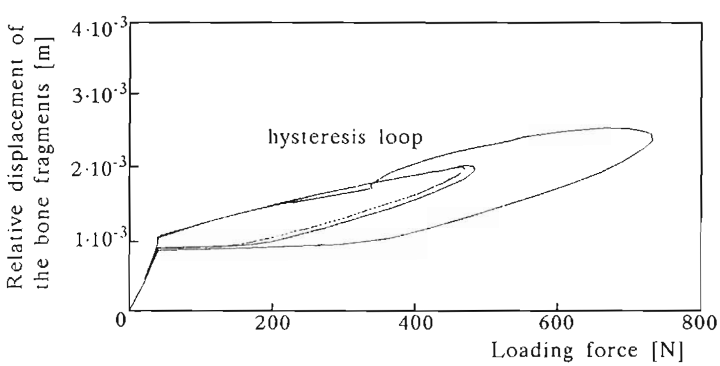

Fig. 13. Simulation results hysteresis loop: the system before the optimising procedure under a variable force

of the dynamisation chamber displacement and the bone-screw strain. It was assumed that the clearance in the dynamisation chamber has been adjusted to ± 1 mm. It can be seen that the real displacements are significantly greater as the component affected by the strain of screws is relatively great. Similar results for varying loads $Q = mg(1-\sin\omega)$ are shown in Fig.12 and Fig.13. The time responses are shown in Fig.12, the hysteresis loop is presented in Fig.13. The bone union dynamisation is too great (with respect to that adjusted by the dynamisation chamber) in this case.

Fig.14, Fig.15 and Fig.16 present similar diagrams for the system after the optimising procedure (executed in accordance with the quality coefficient as defined by the relation (3.16)). The diameters of bone screws were subject to the optimisation procedure. The best diameter obtained was that equal to 6.1 mm. The system performance was excellent. The bone union was not over-stiff, and the micro-displacements did not exceed the admissible values (as adjusted in the dynamisation chamber) in practice.

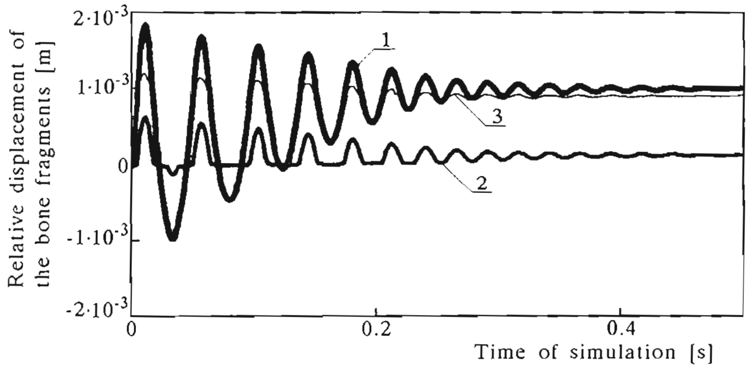

Fig. 14. Simulation results: the system after the optimising procedure under the static vertical force $F_z = 500\,\mathrm{N}$; 1 – total displacement, 2 – component of the bone-screw strain, 3 – component of the dynamisation chamber displacement

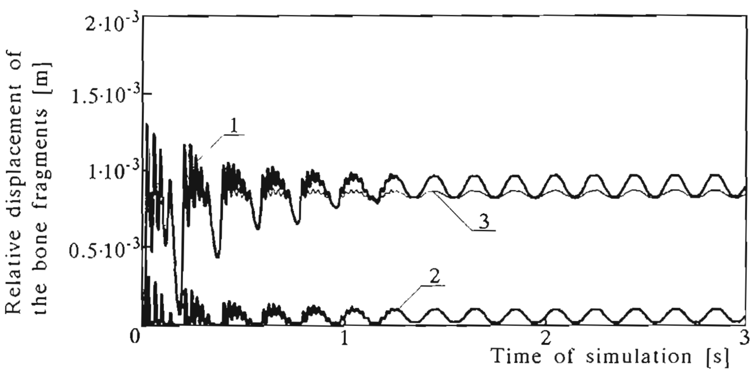

Fig. 15. Simulation results: the system after the optimising procedure under a variable force; 1 – total displacement, 2 – component of the bone-screw strain, 3 – component of the dynamisation chamber displacement

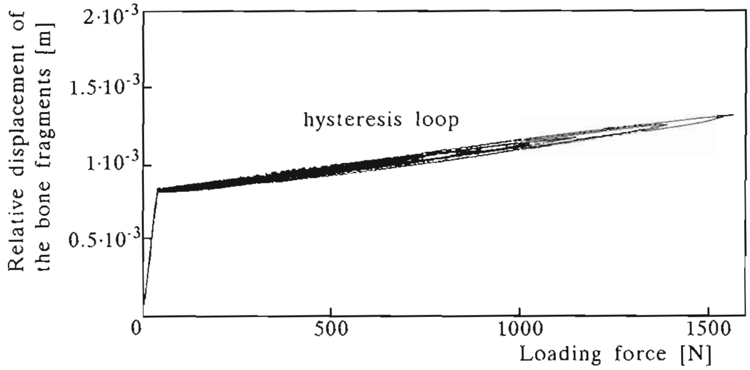

Fig. 16. Simulation results hysteresis loop: the system after the optimising procedure under a variable force

(a) (b)

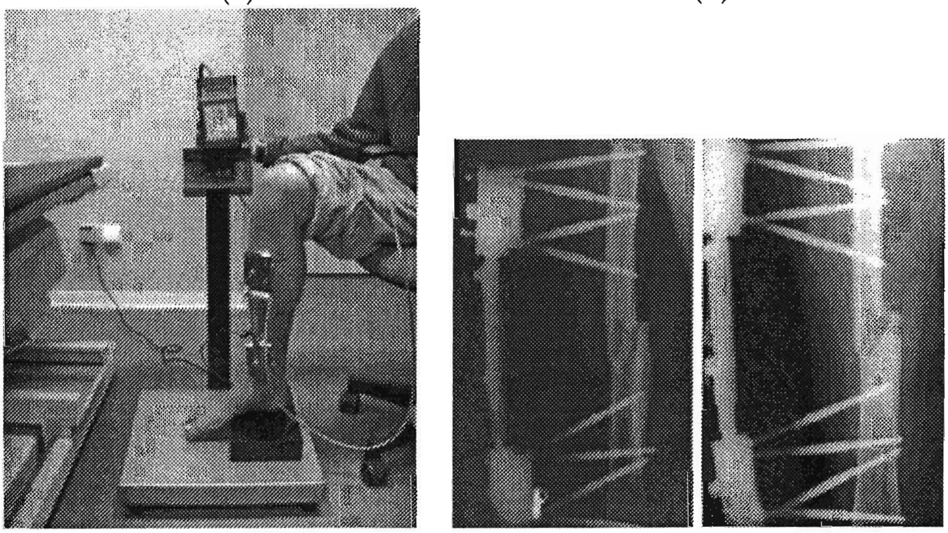

Fig. 17. (a) A patient in the process of determining the bone-union measures (the load F_z is read on the electronic-balance display; the value of the load carried by the bearing frame in the fixator is measured by a strain gauge system and stored in the computer); (b) X-ray picture of the fractured bone with the DYNASTAB DK Fixator installed – the characteristic spatial configuration of bone screws can be seen

5. Clinical verification

The type DYNASTAB DK Fixator designed for healing long-bone shaft fractures had been developed on the basis of the results presented in this paper and was implemented into about 60 bone fracture cases (compound fractures of thigh and lower-limb long-bones) where complications after application of other techniques have been treated. Positive results were obtained in 58 cases, the healing time being reduced by a factor of 18in conventional methods. In author's opinion this resulted – in a great extent – from the postulate of functional treatment of fractures. Clinical pictures are presented in Fig.17, and the diagrams of bone-union measures (those real ones and those from the neurone predictor) are shown in Fig.18. Good agrement between the results can be observed.

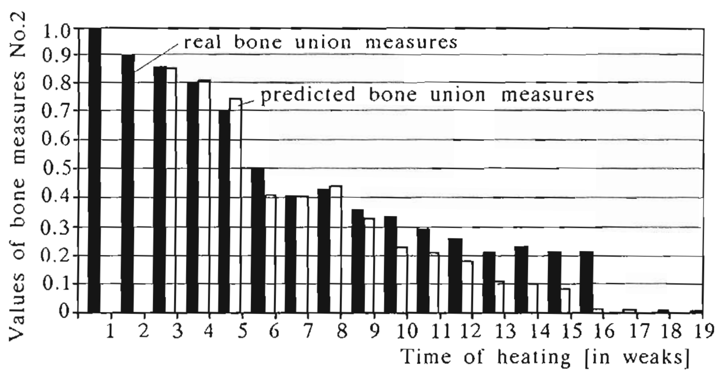

Fig. 18. Diagrams of bone union measures (real ones and those from the neurone predictor)

6. Conclusions

The modelling and simulation techniques presented in this paper were used for designing of mechatronic unilateral external fixators. It appears that the simplified form of the employed discrete model of the external fixator-bone system determines its properties with an accuracy that is sufficient for engineering and medical needs. The term of technical stability is well correlated with the medical term of bone-union dynamisation within the scope of bone-union micro-movements along the axis. A strain gauge test circuit employing an analysing system depending on artificial neurone networks can be employed as an additional and strong diagnostic instrument in clinical practice. For the time being, the clinical experience is promising and proces reliability of the techniques of modelling and simulation.

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Mechatroniczne ortopedyczne stabilizatory zewnętrzne – modelowanie, symulacja komputerowa i badania kliniczne

Streszczenie

Artykuł prezentuje wybrane zagadnienia dotyczące modelowania i symulacji układu: mechatroniczny ortopedyczny unilateralny zewnętrzny stabilizator – kość, przeznaczonego do leczenia trzonów kości długich. Zaprezentowano model nominalny oraz odpowiadający mu model matematyczny, przy czym ograniczono się do modelu dyskretnego. Pokazano związek między klinicznym postulatem czynnościowego leczenia złamań, a pojęciem stateczności technicznej. W artykule zaprezentowano również metodę oceny zrostu kostnego z wykorzystaniem tzw. miar zrostu kostnego oraz metodę symulacji procesów zrostowych wykorzystującą technikę sztucznych sieci neuronowych. Analiza i badania posłużyły do budowy stabilizatora nowej generacji. W pracy przedstawiono również wybrane badania kliniczne.

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