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Modelling of hand implants

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The aim of the research was divided into three problem fields: evaluation of treatment of flexion contractures of interphalangeal joint (two different ways of treatment were compared), optimisation of carpal joint implant (the biomechanical analyses of the behaviour of the original and modified types of the wrist joint implant were carried out) and mechanical behaviour of interphalangeal joint implants (numerical and experimental studies of two deformations states of a finger with the silastic implant were realised for 15° and 75° flexion). The research was performed by hybrid method based on numerical modelling using the Finite Element Method (FEM) by applying ANSYS programme and on experimental investigation using photoelastic measuring.

Keywords: biomechanics, wrist and finger joints implants, numerical modelling, experimental research

1. Introduction

Surgery of the hand has undergone significant development on the world scale, and in recent years also in the Czech Republic. Scientific literature presents new operating as well as therapeutic methods, the origin of a number of new implants and joint replacements. The necessary prerequisite for a successful therapy of such a complex and perfect organ as the human hand consists (in respect of the biomechanical conditions of its function) in the high-quality operating technology and careful post-operative care.

The paper is composed of three parts:

• Evaluation of treatment of flexion contractures of interphalangeal joint: The purpose of the study was to evaluate the stress state inside the joint structure, in other words the articular cartilages, which suffer most during treatment of flexion contractures of the proximal interphalangeal joint (Richtr & Ryšavý, 1992). Flexion contractures are a major therapeutical problem in orthopaedics, plastic and reconstructive surgery of the hand. In most cases the contractures result from profound burns, scald, electric shock, and from secondary healed wounds.

• Optimisation of carpal joint implant: The development of rheumosurgery and surgery of the hand brings about ever newer therapeutical methods. Since 1973 the destructions of the carpal joint and the interphalangeal joints of the hand have been treated by the implantation of silastic endoprostheses of the Swanson type (Swanson, 1973). The Swanson endoprostheses represent progress in medicine, as they make it possible to treat even heavy deformities of the rheumatic hand in a technically relatively simple way (Richtr & Ryšavý, 1993). The implant of the Swanson type consists in a casting, comprising the central transversally flexural zone with proximally and distally projecting slender conical stems. The shape of the implant is determined by the requirements imposed on its function, which is diametrically different from all types of endoprostheses known so far.

The piston effect, i.e. the free movement of the implant stems in the bones, excludes the overloading of the flexural zone. The stems are of rectangular cross sections, which prevents rotation along the longer implant axis. The material of the implant body is silicon rubber based on a methylvinylsiloxan polymer. The material satisfies mechanical requirements and also the requirements imposed on long-term tolerance in human organism.

• Silastic prosthesis of interphalangeal joint: The finger joint implant has similar design as the above-mentioned carpal joint implant. The shape is different in central flexure zone only. In this part a deep groove on the volar side makes easier movement of the wrist.

2. Evaluation of treatment of flexion contractures of interphalangeal joint

We know two ways of treatment: (a) the conservative treatment by gradual splintage, using compressive bandage, (b) traction using a reposition device. The difference in loading between the two above-mentioned methods of joint reconstruction is outlined by Fig. 1.

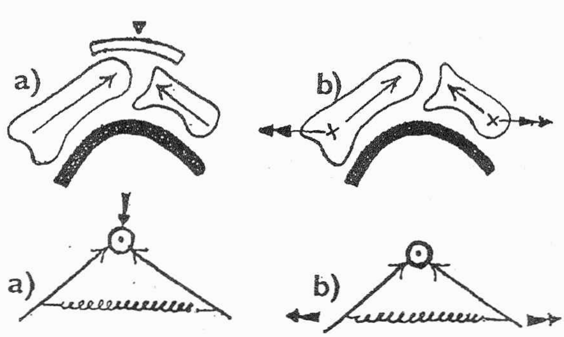

Fig. 1. Scheme of two loading types

2.1. Numerical modelling

The difference in load between the two methods of joint reconstruction is outlined in Figs. 1a–1b. By calculation of the frame structure (Fig. 2), we found out for the same deflection of the joint in the direction of the y axis, i.e. for the same value of the tensile force of the tendon (F = 14.4 N), approximately a triple force acting on the joint in the case of the first "conservative" method (Fig. 2a) in comparison with using a reposition device (Fig. 2b).

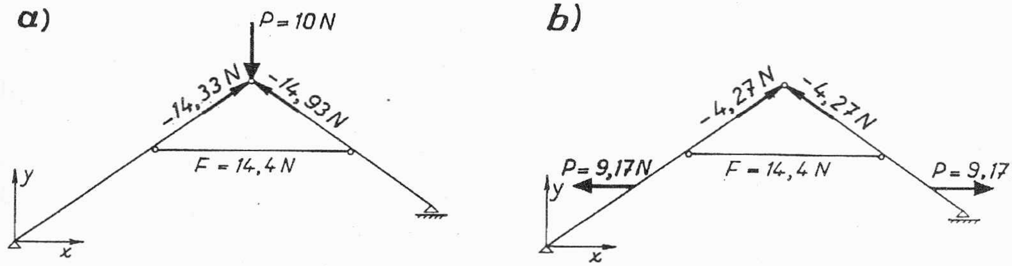

Fig. 2. Frame structure

We substituted the phalanxes for hollow tubes with a modulus of elasticity comparable with cortical bone $E = 17$ GPa and $\mu = 0.25$. Beam elements (hollow tubes) were mutually hinged connected with no regard to cartilage. The tendon was again substituted for a bar hinged connected to both beams with the modulus of elasticity $E = 20$ MPa, which was experimentally measured (Jírová & Kafka, 1993). We considered elastic behaviour of all structures, but we did not calculate with viscoelastic behaviour of tendon, which is in nature significant. It means that the results are correct only for start of loading. The frame structure was calculated as statically determinate structure with fixed bearing on one side and mobile bearing on the other side (Fig. 2). The results have confirmed the experiences of orthopaedic surgeons, who often reported in cases of conservative method of treatment pain and swelling of the joint (Pech et al., 1995). When treatment is accelerated it can result in cartilage necrosis and lamellar chondrolysis (Swanson, 1973).

2.2. Plane FM model

In order to find out the actual stress state of the articulate cartilage it was necessary to design the numerical model of behaviour of the joint structure in the course of loading during treatment. We considered a plane model with the thickness of 1 mm and similar loading was applied as in the case of the frame structure. The model was constructed on the basis of the radiograph of a finger (Fig. 4), when the co-ordinates of nodes were transferred into the computer by using ScanJet IIp. For the analysis we used two-dimensional plane elements (*Plane*) from the programme ANSYS which is based on the Finite Element Method.

The element was used for plane strain analysis of cortical and spongy bone tissues. The soft tissue was substituted for a link (*Link*), no bending of the element was considered. Articular joint was constructed using contact elements, which might be used to represent contact and sliding between two surfaces in two dimensions. The coefficient of dry friction between the articulate cartilages of a joint is very low (it can be as low as 0.0026, nearly the lowest of any known solid material); hence no friction was considered between the cartilage surfaces.

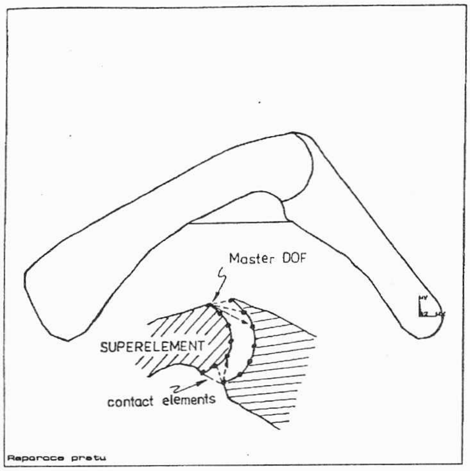

Fig. 3. Finite element model

Because the calculation of a problem is very complicated we have used for the solution of this problem a superelement which was a group of previously assembled ANSYS elements of different types and material properties. The reason for substructuring was to reduce computer time and to allow solution of very large task with limited computer resources. For simplifying solution of a problem we considered spongy bone tissue for both bone structures in the first approximation of our analysis. The task was reduced to one superelement and 1624 contact elements. When we used PC 486 the calculation of this problem took us more than 12 hours even with the use of the superelement.

2.3. Photoelastic measuring

It holds generally, and in biomechanics particularly, that it is always necessary to verify the theoretical model experimentally. If the experimental results prove the correctness of the theoretical model, it will be relatively easier and faster to analyse a number of different loading variants than to devise a new experimental model for each of them.

We have decided to solve the problem using the reflection photoelastic measurement, which appears as more suitable for the ascertainment of stress concentrations in the joint cartilage. We have adopted two different approaches to the manufacture of the model: (i) with reference to similarity in terms of material characteristics, (ii) with reference to similarity in terms of rigidity characteristics.

We prefer the photoelastic measuring because strain gauges provide data only at the points where they have been glued, and little is learned about other regions on the surface of the component. The attention was paid to the qualitative assessment of the places with stress concentrations in the joint structure. The goal was to find out stress peaks and to describe the distribution of stresses during experimental simulation of treatment.

(i) We have modelled a structural system consisting of two phalanges of one material, connected by a cartilage of a different material. The modulus of elasticity of the spongy bone was considered in the FE model as 75 MPa and the modulus of elasticity of the cartilage as 25 MPa. Both materials for experimental research models were made at the same ratio of the moduli of elasticity. Two models were made:

1. The whole model was made of epoxy resins of two different moduli of elasticity.

2. The phalanges were made of aluminium and the cartilage of epoxy resin. The design of this model was based on the fact that we were interested primarily in the cartilage stress state. While keeping the 1:3 ratio of the moduli of elasticity, the epoxy resin modelling the cartilage could have higher modulus of elasticity than that in the first model, as a result of which the material shows pronounced rheological properties which is more favourable for the photoelastic measuring. The material has better optical characteristics enabling more accurate measurements.

(ii) The third model represents similarity in terms of rigidity characteristics. The whole model was made of the same epoxy resin with different thicknesses of the bone and the cartilage.

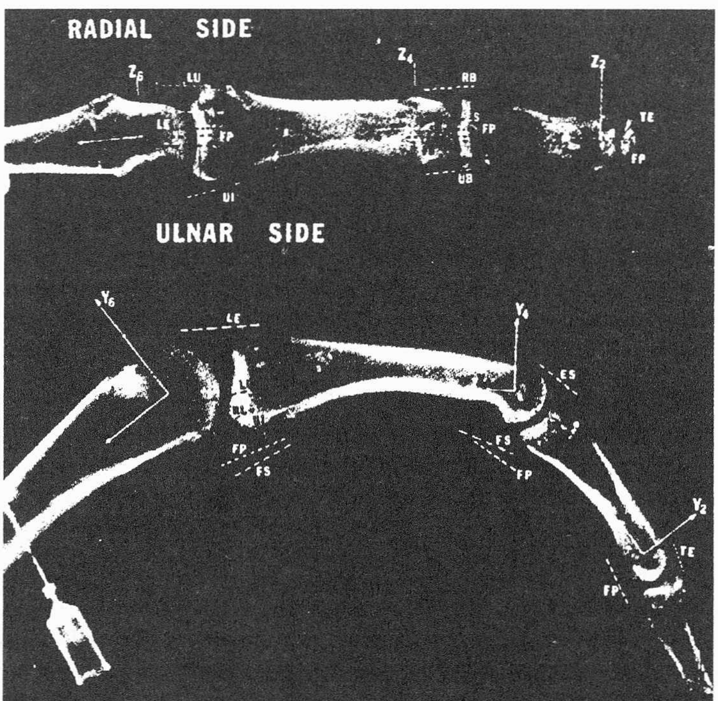

Fig. 4. X-ray views of finger (Chao at al., 1989)

All models were shaped according to the X-ray photo (Fig. 4). The boundary conditions were arranged in accordance with the computational model: fixed hinge at one end, sliding hinge at the other end.

3. Optimization of carpal joint implant

The silastic Swanson wrist joint implant is formed by a one-piece intermedullary silicone rubber casting (Fig. 5) with oval-shaped bending zone and with two tapered rectangular stems. The stems are inserted proximally into the medullary canal of the radial diaphysis and distally through the carpus into the base of the 3rd metacarpus (Fig. 5). A modification of the Swanson endoprosthesis of the wrist by removing from it a 60° wedge at the volar side of the bending zone was proposed by Pech (Pech, 1996). This modification allows for improved movement of the wrist.

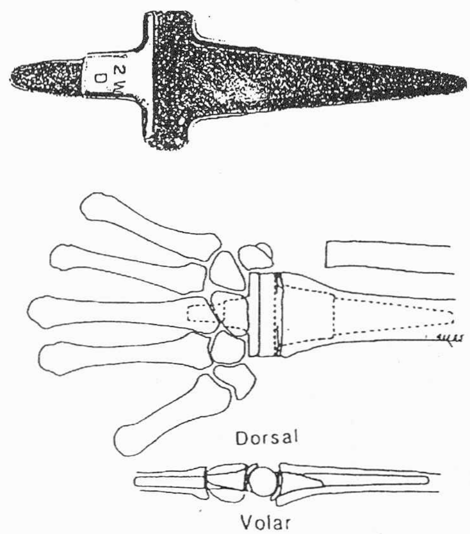

Fig. 5. Silastic Swanson wrist implant (Swanson, 1973)

Two deformation states (Jírová, 1998) obtained by prescribing the displacement of the distal end of the stem implanted in the body of os capitatum and the base of the 3rd metacarpi were analysed: (i) flexion 20°, (ii) extension 30°.

These are the average values of flexion and extension of the postoperative conditions and these degree values were substituted for the displacements of the distal end of the short stem. The fixation of the longer stem in the distal part of radius was simulated by contact elements (Contact 12) that modelled the contact between the implant and the bone along the dorsal and volar sides of the stem. This boundary conditions pattern allows for free movement of the stem in the bone canal along its long axis.

The mathematical model of the original Silastic Swanson Wrist Implant was constructed by using isoparametric elements PLANE 82 (Fig. 6a). This type of element comprises eight nodal points, each with two degrees of freedom. In the two models of the modified wrist replacements the nodes and elements were removed from the place of the grove at the volar side (Fig. 6b) and both at the volar and palmar sides (Fig. 6c) which was done to improve the movement of the wrist. We compared the movement of the original type and the modified ones, E = 100 MPa was considered only as unit modulus of elasticity.

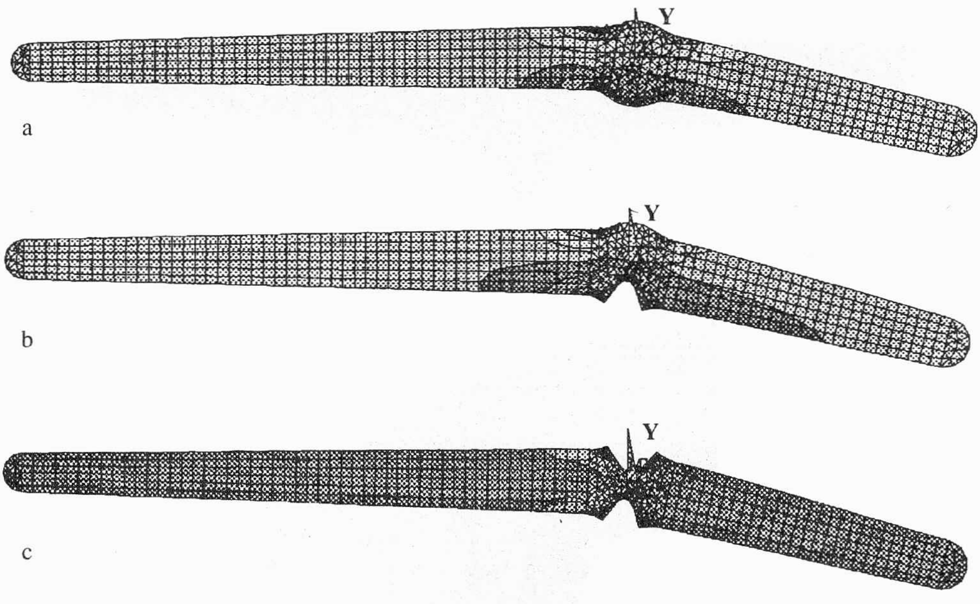

Fig. 6. Models of original (a) and modified (b, c) Swanson wrist implants

4. Silastic prosthesis of interphalangeal joint

The computation was carried out by the FEM using the ANSYS 5.3 programme. The computational model (Jíra et al., 1998) was used for the ascertainment of the stresses, separately on the volar and the dorsal sides, where the greatest stress concentration can be assumed. A three-dimensional model was used with the assumed plane of symmetry of the implant. Type SOLID 95 element from the ANSYS library was used. Two cases of deformation were considered by the prescription of the appropriate rotation in cylindrical coordinates to the distal stem end inserted into the phalangeal bone of the interphalangeal joint: (i) flexure of 15° as the initial case, (ii) flexure of 75° as the maximum case of the function of the interphalangeal joint after the operation.

Practically they are the minimum and the maximum values of flexure of the post-operation state. The fitting of the stem into the proximal part of the radius was simulated by zero displacement of nodal points situated along the dorsal and the volar sides in perpendicular direction with the possibility of any displacement in the direction parallel with the longitudinal axis. The model of the silastic Swanson replacement of the interphalangeal joint consists of 4904 nodes and 2348 type SOLID 95 elements (Fig. 7). This element type is determined by 20 nodes with three degrees of freedom each. For the calculation the modulus of elasticity E = 4.4 MPa was considered.

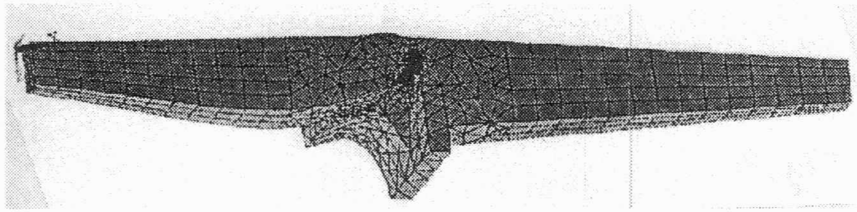

Fig. 7. Computation model of silastic prosthesis of interphalangeal joint

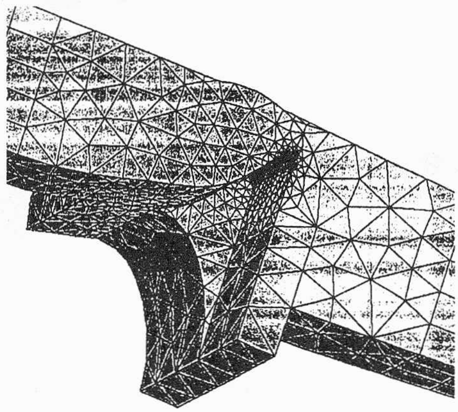

Fig. 8. Detail of bending zone

5. Conclusions and results

Evaluation of treatment of flexion contractures of interphalangeal joint. The performed research has proved that the application of the reflection photoelastic measurements to the investigation of local stress fields in the proximal interphalangeal joint in the course of treatment is sufficiently accurate both qualitatively and quantitatively and not connected with excessive requirements for either time or costs. The first two models yielded qualitatively the similar results as the theoretical model used for numerical analysis (Figs. 5, 6). In the second aluminium and epoxy model the stress concentration was distributed over a larger area. The third model has enabled the evaluation of the stresses not only in the cartilage, but also in the bone tissue of the phalanges.

The whole loading of the articulate cartilage, which has been calculated for the frame structure, is in fact concentrated on a relatively small area, i.e. it is distributed non-uniformly. Experimental research and computational modelling have showed that the articulate cartilage is even considerably unphysiologically overstressed (Figs. 9, 10) during the conservative treatment.

Swanson prosthesis of the carpal joint. Tensional stress concentrations *sig I*, which determine the material load-bearing capacity, are the most important criteria of design evaluation. Since the changed range of movement of the wrist achieved by the modification of the replacement was analysed, the maximum angle achieved by application of the unit load $Fy = 1$ N at end of the short stem was compared. The third quantity assessed was the reaction Ry, i.e. the force developing in the contact area of the resected bone and the endoprosthesis stem during flexion on the volar and during extension on the dorsal sides.

Table 1. Maximum values calculated

Implant type	Loading	Maximum tension (MPa)	Maximum compression (MPa)	Maximum angle (deg) $Fy = 1$ N	Stiffness (%)	Ry (MPa)
Original type	flexion 20°	6.661	–10.217	–9.33	100	–6.289
	extension 30°	10.034	–15.508	9.33	100	–10.357
1st modified	flexion 20°	6. 711	–8.645	–11.93	78	–4.839
	extension 30°	13.061	–12.950	12.24	76	–7.427
2nd modified	flexion 20°	10.603	–10.435	–24.31	38	–3.184
	extension 30°	15.744	–15.810	24.31	38	–4.858

It follows from Table 1 that the modified Swanson wrist implant allows for better movement capability of the operated wrist joint. The tension stress concentrations in the cases of both modified implants were calculated in the top of the cut and the value depends on the cut geometry. But any dangerous tension stress concentration of the modified replacement can easily be eliminated in manufacture.

Silastic prosthesis of interphalangeal joint. The concentrations of tensile stresses (Fig. 11), which are decisive for the ascertainment of the bearing capacity of the given material, are most important for the evaluation. That is why we calculated the values of the principal stress *sig I* versus the developed length of the flexural zone periphery, as it is there that the greatest stresses occur during flexure. The maximum *sig I* values for the implant in Table 2 occur in the same place for both flexure values.

Table 2. Maximum stress of silastic prosthesis of interphalangeal joint

Flexion	Maximum *sig I* (MPa)
15°	0.949
75°	4.738

The evaluation has shown that the Swanson interphalangeal joint replacement, which enables better mobility of operated fingers, has very good mechanical properties. During active use of the hand no major stresses arise in the implant body. Experimentally the replacement can withstand even 250 million flexures. The tensile stress concentration may arise during not very probable extension in the apex of the notch (groove) because of the manufacture of the implant by an adequate technological process.

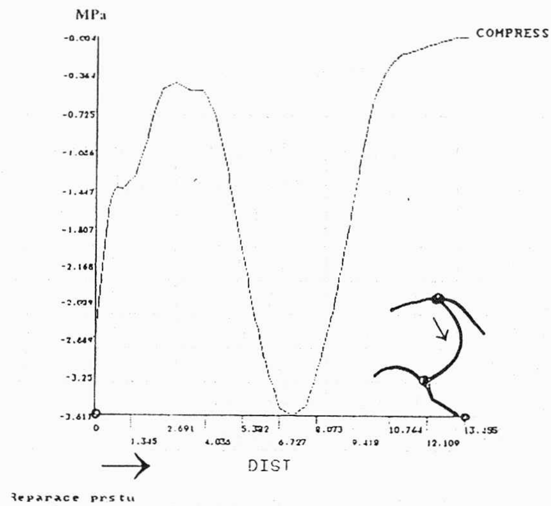

Fig. 9. Numerical results

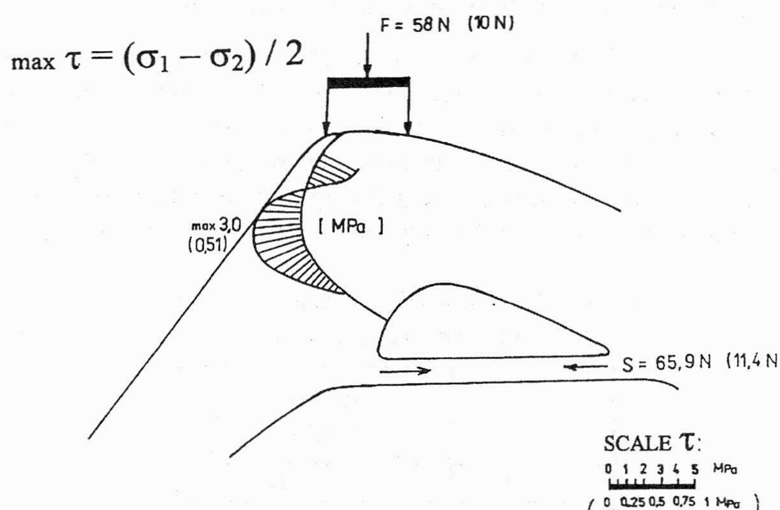

Fig. 10. Photoelastic measuring – stress state distribution

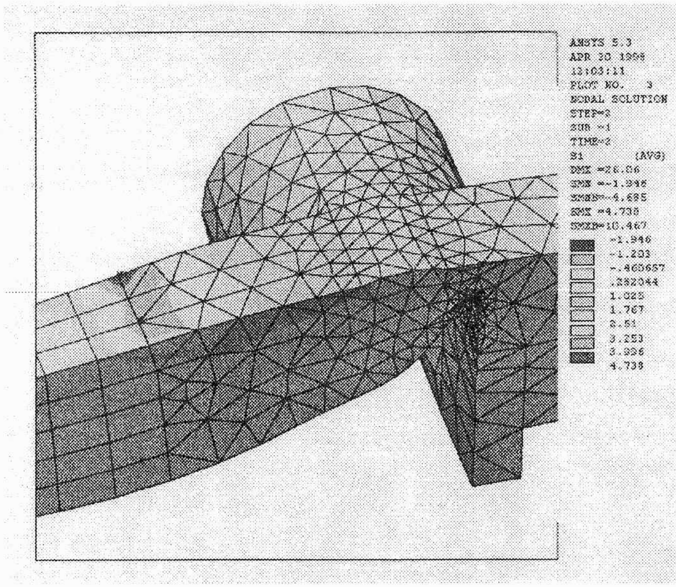

Fig. 11. Stress field *sig I*

Perfect compatibility between the silastic material and easy application of the implants predetermines endoprostheses of the Swanson type for major proliferation in the surgical treatment of the functional apparatus of the hand.

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