Coll. Antropol. **32** (2008) 3: 881–886
Original scientific paper

Biomechanical Model of the Diabetic Foot

Ante Agić[1], Vasilije Nikolić[2], Budimir Mijović [3] and Uwe Reischl[4]

[1] Faculty of Chemical Engineering and Technology, University of Zagreb, Zagreb, Croatia
[2] Faculty of Medicine, University »J. J. Strossmayer«, Osijek, Croatia
[3] Faculty of Textile Technology, University of Zagreb, Zagreb, Croatia
[4] Centres for Health Science, Boise State University, Boise, USA

A B S T R A C T

In this work, a two dimensional (2D) finite element foot model was established from magnetic resonance imaging (MRI) of a male subject. The model comprises first medial planar cross-section through the foot, representing the foot in standing posture. For specified external load, the stress and strain distribution field under foot structure are determined. The material characterization of foot structure components are stronger related to diabetic phenomena. The new material model for soft tissue based on mixture theory is proposed. The linear finite element model replaced by nonlinear counterpart with segment-to-segment contact element.

Key words: *diabetic foot, biomechanical model, finite element method*

Introduction

Diabetes is a growing health problem round the World. Diabetes is a lifelong condition that seriously affects a person's quality of life. The people with diabetes mean probably people with foot ulcers, a common side effect of the disease. It's an unfortunate complication that often leads to amputation. Amputation resulting from diabetes complication is a problem that is difficult to overstate, not just for individuals who have lost limbs, but for the society[1]. Peripheral neuropathy, a loss of feeling in the extremities, renders these individuals unaware of sores that develop on their feet until the wound becomes infected. Then, because of other diabetes-related complications, the infection often defies healing and eventually leads to amputation. One of the most serious complications is naturopathic foot ulceration that, untreated, can lead to lower limb amputation. The main cause of foot ulceration in the adult neuropathy diabetic is thought to be the presence of abnormally high plantar pressures secondary to neuropathy. These pressures may be present as a result of compromised foot function, such as in hind foot tendon disorders and diabetic Charcot foot. The researches believe that with some imaginative techniques and high-tech devices, foot ulcers can be healed, prevented or better say amputation are avoidable[2]. For example, studies have shown that foot morphology (calcaneal pitch angle) affects peak plantar pressure and plantar pressure is related to ulceration. The averted calcanei were associated with medial metatarsal head ulcers, while inverted calcanei were associated with lateral metatarsal head ulcers. Foot deformities, such as hammer/claw toe deformity or hallux limitus, have been significantly associated with ulcer incidence in a univariate analysis. Distribution of the internal foot geometry for given population is indicator of the foot deformity, ageing, and body growth anomaly and many others.

Beside common anthropometric descriptors for the foot there are some a more sensitive measure of foot structure, such as first and fifth metatarsal inclination α, Chopart's joints angle β, inferior calcaneal inclination γ, heel pad thickness δ, first metatarsal thickness h, (see Figure 1). Morphological description of the foot beside geometrical descriptors needs biomechanical factors such as muscle deformation, tissue stiffness. Biomechanical quantificators are important because the foot is always under dynamic loads in variety environments. Recent literature on the diabetic foot indicates that mechanical stress concentrations in deep tissues of the plantar pad of the foot, which develop directly under bony prominences (particularly under the calcareous and metatarsal heads) play a dominant role in the mechanism of diabetic foot injuries and may lead to foot ulceration[3]. There are

Received for publication June 20, 2007

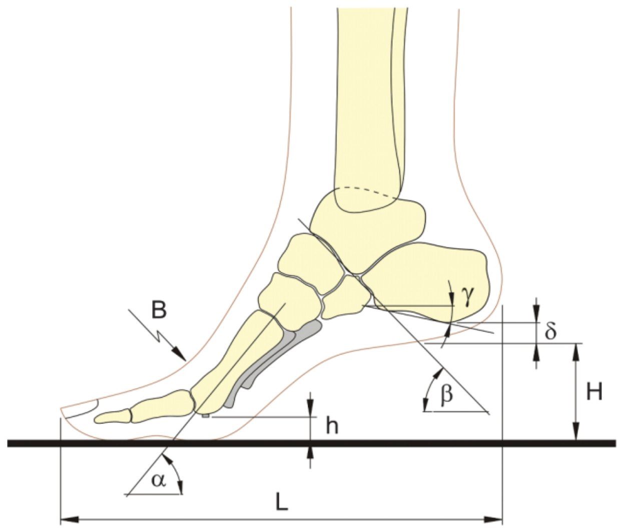

Fig. 1. The foot segments model.

many structural and functional factors that are predictors of foot mechanics phenomena and associated pathology. In research study[4] it is found a strong relationship between foot deformity and ulceration. Foot type was not associated with ulcer development is another conclusion of this study.

The miniaturize sensors can fit inside the shoe in order to measures, records and analyzes a variety of conditions, including pressure, temperature and humidity inside the shoe. Via a radio signal, it provides feedback to a unit which is the patient wears on a body. Eventually, in the shoe can be added audible or pulsing alarms to signal risky conditions. The patient can wear for an extended period of time that will collect information and provide feedback when the patient is at risk for skin breakdown. The »smart« shoe will be useful for anyone who needs ongoing feedback. This device would be like a virtual physical therapist. The objective of this study is to develop a patient-specific biomechanical model of the foot able to predict mechanical stresses transferred through the soft tissues padding the medial metatarsals and calcaneous during standing, and the deformations of these tissues of diabetic patients.

Materials and Methods

Several techniques have been developed to study the morphology, architecture and kinematics of the foot. Researches have used magnetic resonance imaging (MRI), spiral x-ray computed tomography imaging (SXCT) for visualizing and measuring foot structure. The SXCT imaging methods and the pressure analysis are introduced into a biomechanical foot model to quantify the links between internal structures and external pressures on the foot. The first simplified biomechanical foot model is represented as a mechanism with rigid segments connected with kinematics pair's[5]. The advanced biomechanical model of the foot is described in reference[6] where foot multibody dynamics model is extended with flexible-skeletal elements. A biomechanical model of the foot for specified external load gives a stress and strain distribution under foot structure[7]. Measurement muscle and tendons deformations, angular rotations in joints and use inverse dynamics approach provide a valuable insight into the mechanical causes of foot structure behaviour. A modular modelling approach with incorporation of deformable foot segments into multibody dynamics simulation environments (ADAMS, SIMM, and MADYMO) is from greatest importance[8]. Using these models, the researchers hope to determine how the foot responds to stress, allowing for the development of software that would apply computer-assisted design technology to the production of orthotic devices and custom shoes. In the near future we should be able to indicate the optimal characteristics of the insole shape, the location of different kinds of loading patterns, and the material properties[9].

Finite element model

Mathematical models are a promising avenue of investigation complex problem such as diabetes. The computational model (finite element method) of the foot will combine the structure of the foot, the material properties of the all foot components and the pressure on the bottom of the foot[10]. All of the approaches to heal open ulcers, however, are concerned fundamentally with adjusting pressure on the unfeeling foot. If pressure is the key thing that breaks the skin down and keeps the ulcers open, then if we can reduce the pressures they should be able to heal very quickly. In this paper biomechanical foot model comprises first medial planar cross-section through the foot, representing the foot in standing posture. The geometric data of the foot's skeletal cross-section of an adult male were scanned using magnetic resonance imaging (MRI). The foot shape of the subject was obtained from surface digitalization via a 3D laser scanner. The foot geometry is transferred to a commercial finite element analysis program GID[11] adopted for this problem, in order to construct multibody flexible structural model. The foot model is discretized into isoparametric finite element mesh representing the bones, cartilages, ligaments and soft tissue. The finite element model of the human foot is shown by Figure 2. Every part of the structure is meshed as separate region in order to give material attributes for every element. The total load carried by the foot model was determined to be 350 N, which were distributed as 25–19–19–19–18 % for the first through the fifth rays, respectively[12]. The reaction of the Achilles tendon force is approximated as 50 % of the body load during standing. Supports constraining the model's vertical displacements were positioned under the hell base and under the metatarsal heads. The foot-ground interface is assumed as frictional contact with coefficient of friction as 0.5[12].

The equilibrium equation

The equation of motion expressed by Lagrange equation of the foot structure has the following form[8]

$$\frac{d}{dt}\left(\frac{\partial T}{\partial \dot{q}_i}\right)+\frac{\partial V}{\partial q_1}=Q_i \qquad i=1,2,..n \tag{1}$$

where T and V are kinetic and potential energy, respectively. The q_i and Q_i are components generalized displacement and force vector respectively, and n is a number degree of freedom. The equation of static equilibrium for the foot as flexible-rigid multibody system is described by matrix equation

$$[K]\cdot\{q\}=\{\mathrm{F}\} \tag{2}$$

where $[K]$ is the stiffness matrix, $\{q\}$ is nodal displacement vector, and $\{F\}$ external force vector. The degree of freedom can be partitioned into those of these interface or boundary nodes u_B and those of the interior nodes u_I. The basic equation of force balance can be written in partitioned form

$$\begin{bmatrix} K_{BB} & K_{BI} \\ K_{IB} & K_{II} \end{bmatrix}\begin{Bmatrix} u_B \\ u_I \end{Bmatrix}=\begin{Bmatrix} F_B \\ F_I \end{Bmatrix} \tag{3}$$

The constraint modes can be computed by solving the lower partition of above equation as follow

$$u_I=-K_{II}^{-1}\cdot K_{IB}u_B+K_{II}^{-1}F_I \tag{4}$$

The deflection of all the interior degree of freedom is expressed as sum of two parts; the boundary degree of freedom and interior loading part.

The matrix expressions for strain-displacement, and stress-strain, for the 2-D model involving finite elements are given in classical textbook for finite element method and therefore not discussed here[13]. The stress analyses are carried out using the commercial software, (GID, CIMNE, Barcelona)[11] modified for biomechanics related phenomena. In order to simulate the surface interactions among metatarsals, cuneiforms, cuboids, navicular, talus and calcaneus's surface-to-surface contact algorithm was used, which allow relative movement. Contact procedure use a segment-to-segment contact approach based on polygonal contact model[14]. The global variational principle for the two-body contact system has the following form

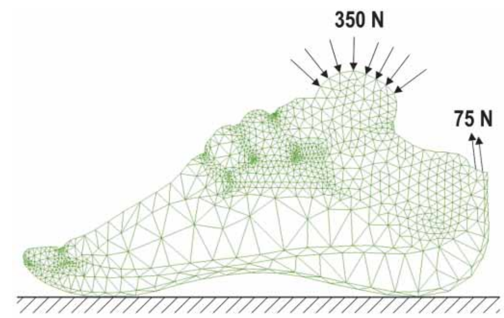

Fig. 2. Foot finite element model.

$$\int_V \sigma_{ij}\cdot\delta\varepsilon_{ij}=\int_V f_v\cdot\delta u\, dV+\int_\Gamma f_\Gamma\cdot\delta u\, d\Gamma+\int_\Lambda t_\Lambda(\delta u_1-\delta u_2)d\Lambda=0 \tag{5}$$

t_Λ is the contact traction on surface Λ, δu_1 and δu_2 are vectors of contact surfaces nodal coordinates. The vectors f_v and f_Γ are externally applied body and surface force vectors. The σ_{ij} and ε_{ij} are stress and strain tensors, respectively. For this study a simple master-slave approach was implemented whereby contact checking points are placed at slave surface integration points rather than nodes, suitable for the cubic Hermite surface used in our software. We are used concept of thin elastic layers between contacting surfaces[14]. In other words, discretization procedure influence of elastic layers con-

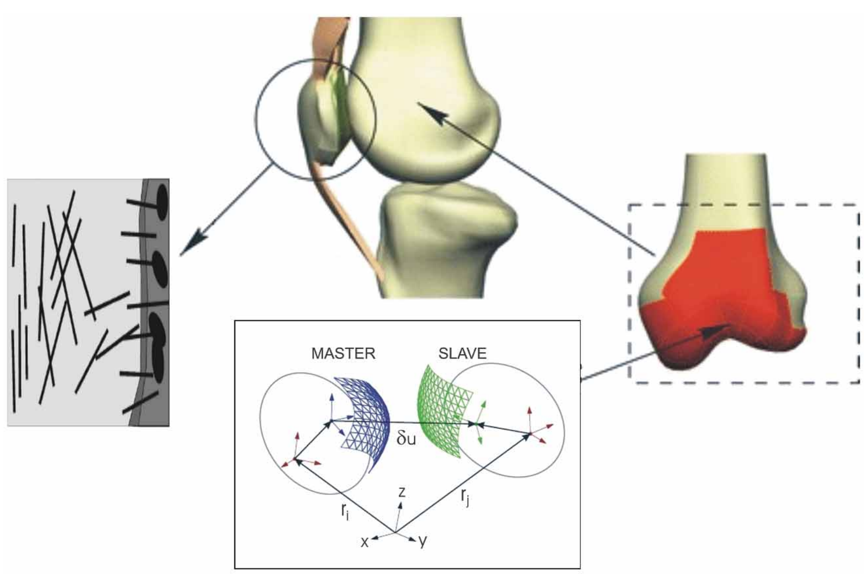

Fig. 3. The segment-to-segment contact.

verts in a set of springs scattered over the surface of bodies.

Normal forces of a contact element can be written as

$$F_n = c \cdot A \cdot u_n \tag{6}$$

where is u_n penetration vector in normal direction $\vec{n}$, A is the area of the contact element and c is the layers stiffness. The layer stiffness (elastic foundation model) is defined by

$$c = \frac{1-\nu}{(1-\nu)\cdot(1-2\nu)}\frac{E}{h} \tag{7}$$

where h is layer thickness, E and ν are layers Young's modulus and Poisson's ratio respectively.

Material properties

Material characterization of the diabetic tissues had influenced should by chemical dynamics processes specific for diabetic body. The material model philosophy should be able to distinguish difference between diabetic and healthy tissue. It is well-known the material properties diabetic and non-diabetic person are different for other organ too. A multiscale material model is proposed in work[15], based on bio-chemo-mechanical foundations. According thermodynamics of mixture theory, we proposed the following expression for the free energy of the diabetic person soft tissue

$$\psi = \psi\,(I_1, I_2, .. I_n, \vartheta, c_1, .. c_m) \tag{8}$$

where $I_1, .. I_n$ are invariants of the deformation tensor and elastic microstructure descriptor tensors, ϑ is temperature, and $c_1, .. c_m$ are internal chemical coordinates (glucose and insulin concentration, receptor and cell descriptors, cross-link density). For example, cross linking of collagenous tissue link mechanical properties with physiological states in the foot[16]. Therefore, fourth-order elastic tensor for the tissue has the following form

$$\Xi_{ijkl}(c_1,..,c_m) = \sum_n \sum_m \left(\frac{\partial^2\psi}{\partial I_n \partial I_m}\frac{\partial I_n}{\partial C_{kl}} \otimes \frac{\partial I_m}{\partial C_{ij}} + \frac{\partial\psi}{\partial I_n}\frac{\partial^2 I_n}{\partial C_{ij}\partial C_{kl}} \right) \tag{9}$$

where C_{ij} is Cauchy-Green deformation tensor. The connection between the multiple materials scales are overcome by stochastic homogenization procedures. Bone and cartilage material's properties were assigned as having the linear, elastic and isotropic properties, while soft tissue, fat pad, ligaments, fascia were considered as being non-linear materials. (Table 1). Second-order polynomial strain energy potential was used for soft tissue[17]

$$U = C_{10}(I_1-3) + C_{01}(I_2-3) + C_{20}(I_1-3)^2 + C_{02}(I_2-3)^2 + C_{11}(I_1-3)\cdot(I_2-3) + D\cdot(I_3-1)^2 \tag{10}$$

where are I_i, i=1,2,3 are strain invariant, and C_{kl}, k,l= 0,1,2 and D are material constants respectively. The stress-strain behaviour for ligament and fascia are expressed in polynomial form[12]

$$\sigma = A_0 + A_1\cdot\lambda + A_2\cdot\lambda^2 + A_3\cdot\lambda^3 + A_4\cdot\lambda^4 + A_5\cdot\lambda^5 \tag{11}$$

where A_i are material constants, and λ is stretch ratio.

TABLE 1
MATERIAL CONSTANTS

C_{10}	C_{01}	C_{20}	C_{02}	C_{11}	D
0.085	–0.058	0.039	0.0085	–0.023	3.65

TABLE 2
MATERIAL PROPERTIES

	Modulus MPa	Poisson ratio
Ligaments	200	0.4
Soft tissue	hyperelastic	0.5
Tendon	15	0.1
Cartilage	1	0.4
Skin	hyperelastic	0.5
Fascia		0.4
Bone	7300	0.3

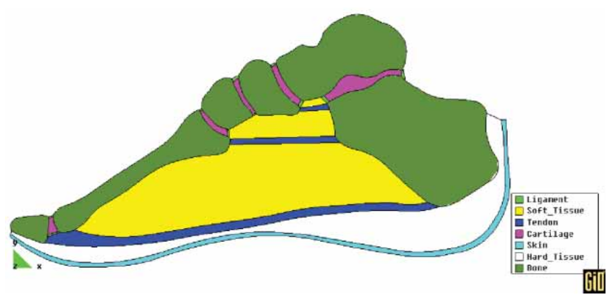

Fig. 4. Foot material model.

Results and Discussion

An anatomically detailed 2D finite element model of the human foot complex was developed from CT scan images and CAD manipulation. Kinematics constraints between bone structure, cartilage and soft tissues were defined by contact elements. There are two mechanisms for transfer compressive load during the standing posture:

- stretching of the plantar fascia under the load
- bending of the metatarsal bones

Distribution horizontal displacements are shown by Figure 5. The stretching of the plantar fascia is evident. The compressive deformation near the metatarsal head

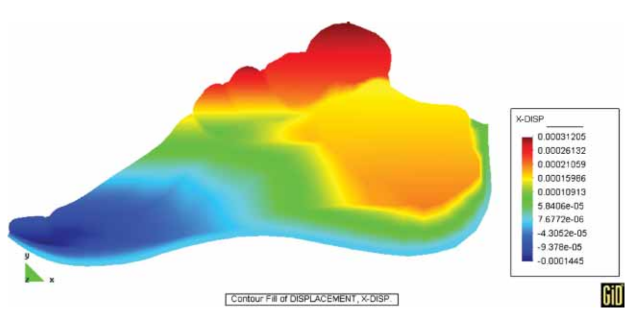

Fig. 5. The horizontal displacement distribution.

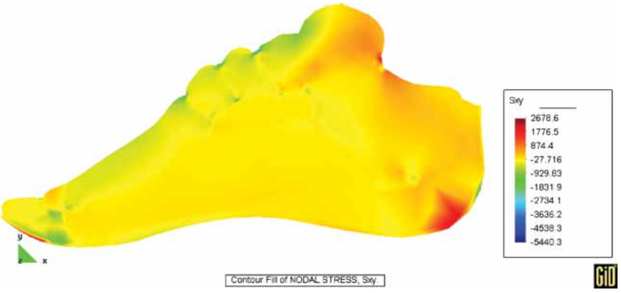

Fig. 6. Shear stress distributions.

indicate the possible a place of the stress concentration. The place of stress concentration needs special attention, because this is probably a place of ulcerations. Many researches believe that neuropathic ulcers begin internally and progress to the skin surface[18].

The distribution of the shear stresses is shown by Figure 6. The change of sign of the shear stresses is visible on interface between bone and tissue. The significant tension stress concentration in plantar skin and deep soft tissues under the metatarsal heads occur, which do not appear in the normal foot model.

The biomechanical model of the foot is really complex structural problem. The multibody dynamic musculoskeletal systems need special element such as contact and interface elements. The material properties discontinuity at interface is another problem. In this work, a preliminary 2D finite element foot model was established using magnetic resonance imaging (MRI) of a male subject. The model replaces three-dimensional foot structure with planar cross-section model, which validation is open for discussion. The preliminary result indicates agreement stress and deformation patterns with literature data[12,17]. Models reported in literature differ with respect to the number of segments defined, the number of dimensions considered, and the algorithms used to solve the stress-strain problem. Additionally, the anatomical information and the material properties used in the different models are not identical. Consequently, it is difficult to compare the results of different models directly. The bulk soft tissue stiffness increase up to 5 times the normal values was used to characterize the tissue hardening behaviour under increasing stages of diabetic's neuropathy. The results showed that increasing soft tissue stiffness led to a decrease in the contact area between the plantar foot and the support surface with increases in peak plantar pressure at the forefoot and rear foot regions. The next important structural element was the plantar fascia modulus. The plantar fascia stiffness decrease reduce the arch height, increase the strains of the plantar and spring ligaments. Plantar fascia release increased the strains of the plantar ligaments and stress in the metatarsal bones. Therefore we can suggest only partial surgical release of the plantar fascia, if necessary, in order to minimize the effect on its structural integrity.

Footwear construction example

Design and selection of a suitable shoe for a diabetic patient is mostly based on the experience and intuition of the shoemakers rather than on scientifically based design principles. The footwear design intervention should be made in order to protect the foot at sites that are at risk for plantar ulceration or re-ulceration by reducing pressure to a level below some threshold for ulceration. A variety insole design principles are frequently used to relieve peak plantar pressures[19]. The running shoe bottom cross-section construction is used as brilliant example of systematic reduction contact plantar pressure by bottom design. The wide range of patient characteristics and too many design variables are reason for absence generalized design criteria. A sensitivity analysis of design variables by computer simulation is another guiding design strategy in the prescription of therapeutic footwear. A plane strain bottom model was created using finite element method (see Figure 7). The principal mechanism for contact pressure p_1 reduction may be its distribution over a large area. In the bottom design is introduced airbag which act as air spring with distributed pressure p_3. The bottom with airbag and air channels acts as shock isolator and with pneumatic damping element. Other orthotic and supportive device can be used in order to assure re-distribution contact pressure.

On Figure 8 is shown distribution of von Mises equivalent stresses in bottom cross-section. The use layered

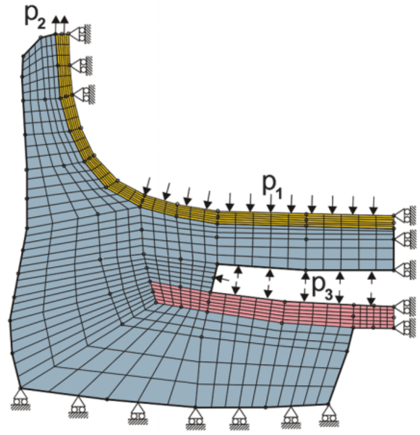

Fig. 7. Footwear bottom finite element model.

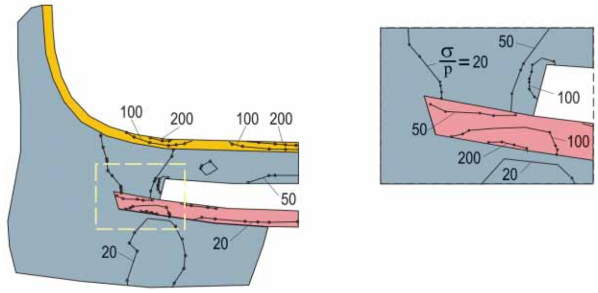

Fig. 8. The von Mises stress distribution.

materials we can control spread of stress magnitude near foot insole contact in order to prevent stress concentration.

Conclusion

In this paper we are presented diabetic foot modelling procedure. Finite element analysis of internal stresses shows great promise as being an effective tool in treatment planning. In order to make difference between diabetic and healthy person material description are improved with new material description on micro level and other degenerative changes on macro scale. The biomechanical model for diabetic foot must be extended on dynamic methodology in order to form patient-specific, anatomically accurate model of the foot, for clinical prognosis, be an aid in prescription of special footwear, simulation of consequence of surgical procedure.

REFERENCES

1. HANČEVIĆ J, COCE F, BOŽIKOV V, Dijabetičko stopalo. In Croat. (Medicinska naklada, 2006, Zagreb). — 2. Consensus Panel, Diabetes Care, 27 (2004) 2262. — 3. MORAG E, CAVANAGH PR, J Biomechanics, 32 (1999) 359. — 4. LEDOUX WR, SHOFER J, SMITH, G. SULLIVAN K, HAYES SG, ASSAL M, REIBER GE, J. Rehabilitation Research & Development, 42 (2005) 665. — 5. SCOTT SH, WINTER DA, J Biomechanics, 26 (1993) 1091. — 6. AGIĆ A, Coll Antropol, 30 (2006) 815. — 7. SADEGHI H, ALLARD P, BARBIER F, SADEGHI S, HINSE S, PERRAULT R, LABELLE H, The Knee, 9 (2002) 47. — 8. WASFY TM, NOOR AK, Appl. Mech. Revue, 56 (2003) 553. — 9. CHEUNG JTM, ZHANG M, Medical Engineering & Physics, 30 (2007) 269. — 10. LIN CS, 3D Dynamic FEA of a Normal Foot, MS Theses. In Chinese (Chung Yuan Chris. University, Taiwan, 2004). — 11. GID – Users Manual (CIMNE, Barcelona). — 12. GEFEN A, Medical Engineering & Physics, 25 (2003) 491. — 13. ZIENKIEWICZ OC, The finite element method (McGraw Hill, London, 1977). — 14. EBRAHIMI S, HIPPMANN G,EBERHARD P, Int J Appl Math and Mechanics, 1 (2005) 33. — 15. AGIC A, TRNSKY D, DeSYO D, Ergonomics 2007, (3th Int. Ergonomics Conference, St. Toplice, 2007). 33. — 16. LAYTON BE, SASTRY AM, J. Biomechanical Enginering, 126 (2004) 803. — 17. CHEUNG JTM, ZHANG M, ABAQUS Users' Conference 2006. — 18. ACTIS RL,VENTURA LB, SMITH KE, Med. Bio Engineering Computation, 44 (2006) 653. — 19. GOSKE S, ERDEMIR A, PETRE M, BUDHABHATTI S, CAVANAGH PR, J Biomechanics, 39 (2007) 2363.

A. Agić

Faculty of Chemical Engineering and Technology, University of Zagreb, Marulićev trg 19, 10000 Zagreb, Croatia
e-mail: aagic@marie.fkit.hr

BIOMEHANIČKI MODEL DIJABETIČKOG STOPALA

S A Ž E T A K

Dvodimenzionalni model stopala s konačnim elementima načinjen je na temelju slike magnetske rezonancije muške stope. Model predstavlja prvi medijalni poprečni presjek stope u stojem položaju. Za zadano vanjsko opterećenje određen je raspored naprezanja i deformacija unutar stopala. Materijalna karakterizacija segmenata stope je usko povezana s fenomenima dijabetesa. Za meko tkivo predložen je novi model materijala baziran na višekomponentnoj termodinamici materijala. Linearno elastični model konačnih elemenata zamijenjen je nelinearnim modelom s kontaktnim algoritmom.