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A VISCOELASTIC MODEL FOR SKIN OF STUMP IN TRANSTIBIAL PROSTHESIS

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Abstract: *The displacement of different parts during gait cycle as force and pressure effect can contribute to developing and increasing of lesions. In present paper we investigate the viscoelastic model for prosthetic skin using a rheological model, the Maxwell slip model. We propose a Maxwell model for hysteretic behavior of skin (force vs. displacement) at the interface of stump-socket in transtibial prosthetic case. Various methods for parameter optimization in the Maxwell model are discussed and a novel method is proposed: genetic algorithm with multiple separated chromosomes in order to find the minimal number of elasto-elements with maximum performance of model. The Force and displacements are measured using Finite element model (FEM) and simulations. The experimental results of simulation take into account two different form of hysteresis for two different locations on stump.*

Key words: *Hysteresis, Maxwell model, genetic algorithms, stump-liner and stump socket interface, transtibial prosthesis*, finite element model.

1. INTRODUCTION

The interface of the pressures of blunt-liner has a distinctive and important achievement of a comfortable prosthesis for a person with the lower level limb amputation. An amputation is a surgical sacrifice of a section of anatomic parts: bones, muscles, blood vessels and nerves.

Tribology at the level of skin plays a very important role in patient's comfort. If the friction coefficient is too high, the pressure due to load in the cycle gait could cause damages at the tissue level and even depth injuries in the stump. Counterpart, a low friction coefficient can cause the slip of the socket out of stump and this problem can lead to serious injuries or major patient's discomfort. A tradeoff between these two requirements is practical a multiobjective optimization problem that requires a feedback form patient about the optimum of this comfort. The most common evaluation of patient's comfort is Prosthetic Evaluation Questionnaire (PEQ) that uses a scale from 1 to 5 in order to evaluate the qualitative comfort, where 5 is the greatest level [1].

If the construction of prosthesis will not be correlated with the training of the residual for functioning, especially after the definitive prosthesis is placed, the program of recovery of strength and muscle tone will not be continued, and the blunt prosthesis system will be altered. Once the tissues are destroyed, practically it is no longer possible to use the prosthesis and a new surgical intervention in order to repair or change the blunt will be needed. In order to avoid these changes, the prosthesis must be designed in such a way so that the sensitive areas of the amputated limb would bear a smaller pressure on the part of the prosthesis and the socket, while the more resilient regions would bear a higher pressure.

The displacement of different parts during gait cycle as force and pressure effect can contribute to developing and increasing of lesions. The characteristics of the individual skin have an influence over the resistance of skin, that is, the skin can be more resilient or not, and the model used for skin representation should take into account these particularities.

Literature in tribological behavior of skin and especially scar skin is reduced to few papers well documented and the data provided by these papers are relatively scarce. An interesting paper is that deal with tribological aspect of scar skin in prosthetic application is [2]. The authors used for experiments an MT-II Micro-Tribometer in order to simulate the rubber between skin and prosthesis for three kinds of skin. The sensation experienced by the patients is translated into four levels of qualitative opinion: Normal, Slight, Marked and Severe. The shape of curves force and displacement are different when hysteresis is constructed, that is the force is represented as function of displacement (the variable time is eliminated from equation for force and displacement).

Tribology of healthy skin by quantitative experimental analysis is studied in [3]. Dry skin and moisten skin showed a different coefficient of friction depending also of different anatomical parts and different measurement techniques. The friction coefficient as function of

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the contact pressure showed a large dispersion in calculation based on experimental results, suggesting the adhesion friction model can explain this results [3].

In [4], OLS 1100 microscopy was used to in order to see the microscopic changes in the skin under friction. Mechanical friction, comfort sensation, scoring of injuries (erythema and edema) based visual evaluation are also made [4]. Four socks fabric in evaluation of friction of skin against textiles.

The relation between elastic modulus of skin and surface area of contact is essential for rules that define the design of surfaces [5]. A complete mechanical model of skin must take into account all the characteristics of the skin: anisotropic, nonlinear, and viscoelastic. When the length of scale increases, the elastic modulus decrease also with few orders of magnitude [5].

Friction at human skin level is investigated in [6]. The main objective of the paper is to present a new portable tribometer that is used to investigate the friction between skin and different materials [6].

A 2D finite element model for transfemoral prosthetic stump including skin in proposed in [7]. The friction/slip conditions between skin and socket are simulated using interface elements [7]. The influence of load transfer is studied and the graphics that results from simulations (pressure distribution, shear stress, slip values versus different coefficient of frictions) are showed [7].

In present paper we investigate the viscoelastic model for prosthetic skin using a rheological model, the Maxwell slip model.

2. HYSTERESIS IN TRANSTIBIAL PROSTHETIC MODEL

The most important source of injuries of skin in prosthetic interface is due friction between skin and socket. Two objects in frictional contact will always produce a displacement due to tangential forces depending of contact stiff. During the gait cycle, the local points are subject of cyclic load force that produces local displacement of skin, Fig. 1.

One of the most plausible models of the skin is based on viscoelastic property of it. Viscoelastic property is often associated in mechanics with combination of spring and dampers in different connection schema, but also some viscoelastic models based of friction schemas can be constructed. The force that acts in different points on the stump has various forms Fig. 2.

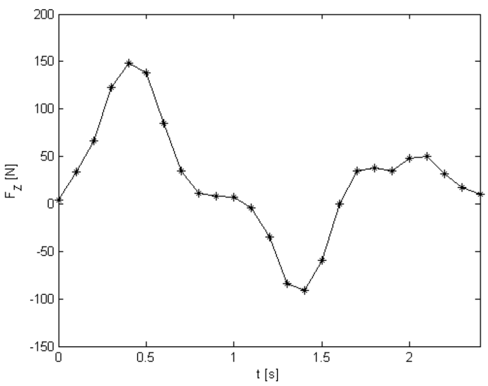

Fig. 1. F_z force during 2.4 sec. cycle load.

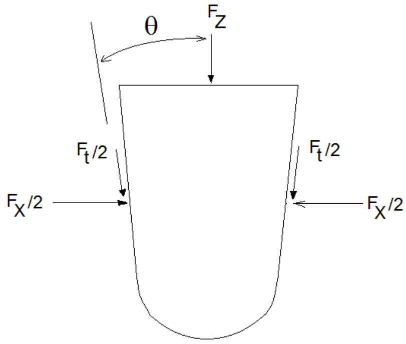

Fig. 2. Load forces on the residual limb.

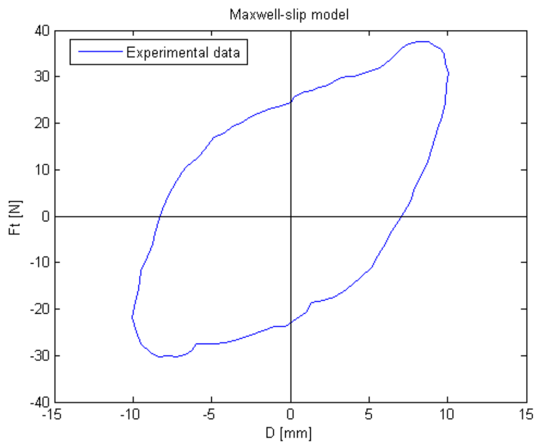

Fig. 3. F_t-D curve, medial location.

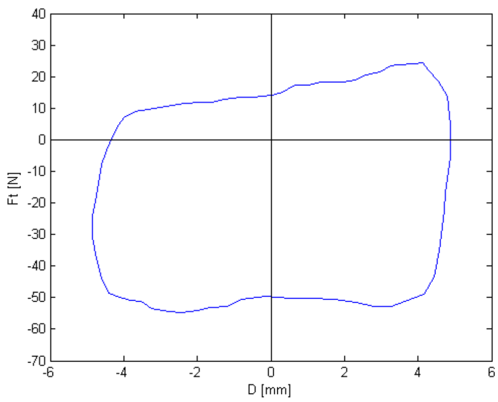

Fig. 4. F_t-D curve, medial location.

The curves that form the hysteresis are made from two curves f(t) = $F_t(t)$ and g(t) = $D(t)$ by eliminating the variable t (time). Depend on location, and load (determined basically by patient's weight but also the correctness of patient's walk) the hysteresis shape can have different forms, Figs. 3–4. The geometry of the stump along with residual bones (tibia and fibula) plays an important role in hysteresis curve.

The coefficient of friction depend basically of the materials used in construction of liner/socket by also it is a particular dependence of intrinsic patient's skin properties. The pre-pressure is needed in order to prevent slip when the maximum vertical load is supported by stump (a typical value is around 800 N, depending on the patient).

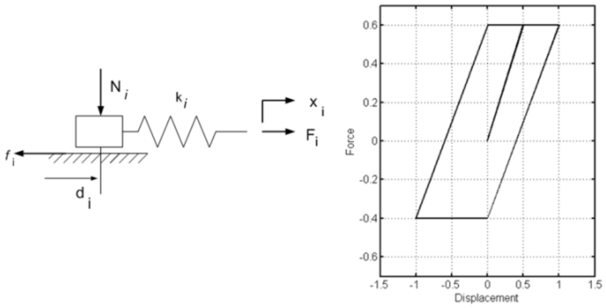

Fig. 5. A single elasto-slide element and its model.

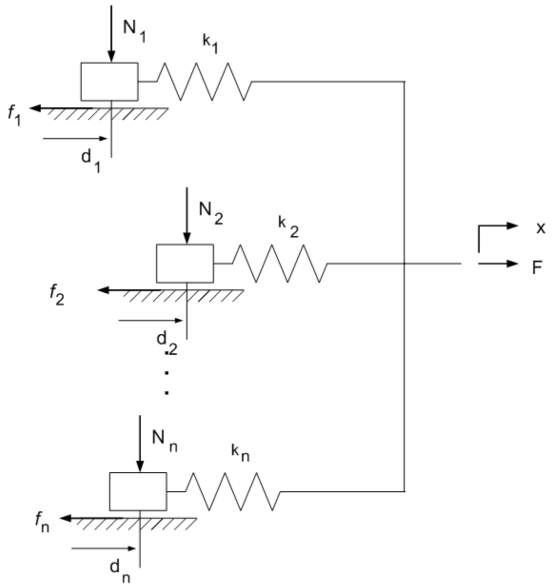

Fig. 6. The force-displacement variables in Maxwell slip model.

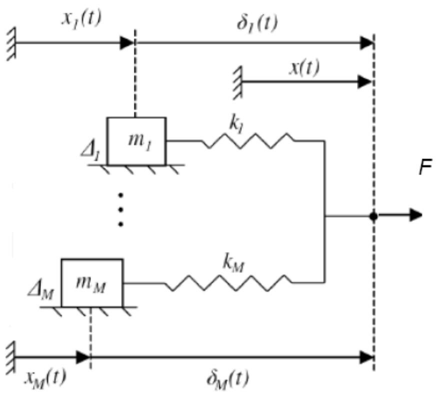

Fig. 7. F_t – The Maxwell slip model used for concatenated formula.

The skin modeling can be useful in modern prosthesis that uses a control of pressure depending on load characteristics. Adaptive control engineering based on plant model is a suitable one for this situation. Controlling force, the displacement is calculated and it can be reduced or increased in order to prevent lesion of skin (fractures, like ragadaes).

Simulations can be used for determination of D-F_t curve in order to use a mathematical model for hysteresis.

3. MAXWELL MODEL FOR SKIN UNDER CYCLIC LOAD

A hysteresis function with nonlocal memory can be modeled by a superposition of many elementary components. Generally, the hysteresis is characterized by variables related by a non-single value function relationship. One of the most common representations is the relationship between force and displacement.

The model is defined by n elasto-elements in parallel, Figs. 5–7. The each elasto-element i has the same input x and one output f_i and it is characterized by a maximum force, a linear spring constant k_i and a state variable x_i describing the position of element i. The behavior of each element is described by:

$$F_i = \begin{cases} k_i(x-d_i) \; if \; |k_i(x-d_i)| < f_i \\ f_i \operatorname{sgn}(x-d_i) \; and \; d_i = x - \dfrac{f_i}{k_i}\operatorname{sgn}(x-d_i) \quad else \end{cases} \tag{1}$$

$$F = \sum_{i=1}^{n} F_i \; . \tag{2}$$

The presliding friction modeling is practically a weighted superposition of M piecewise linear massless operators. In practical implementation we will apply a more convenient formula, given by [7]:

$$\delta_i(t+1) = \operatorname{sgn}[x(t+1) - x(t) + \delta_i(t)] \cdot \min\{|x(t+1) - x(t) + \delta_i(t)|, \Delta_i\}. \tag{3}$$

$$F(t,k,\Delta,D) = \sum_{i=1}^{M} k_i \cdot \delta_i \; , \tag{4}$$

$$k = [k_1, k_2, \ldots, k_M]^T \; , \tag{5}$$

$$\Delta = [\Delta_1, \Delta_2, \ldots, \Delta_M]^T \; , \tag{6}$$

$$D = [\delta_1, \delta_2, \ldots, \delta_M]^T \; . \tag{7}$$

If n goes to infinite we have a generalized Maxwell slip model. In order to identify the model parameters, several methods have been proposed: the linear regression (LR), dynamic linear regression (DLR), the nonlinear regression (NLR), NARX models, neural networks, local models, dynamical networks [8].

Two methods were chosen, the LR that is very simple and a proposed methods the uses genetic algorithms, more complex but that can have the advantage to optimize all the parameters simultaneously.

4. GENETIC ALGORITHM WITH MULTIPLE SEPARATED CHROMOSOMES (GA-MSC) FOR GLOBAL OPTIMIZATION OF PARAMETERS IN MAXWEL MODEL

Genetic algorithms are powerful tools for single objective or multiobjective optimizations [9–10]. We propose to use novel algorithm, genetic algorithm with multiple separated chromosomes (GA-MSC) in order to optimize both K and D parameters. There are few papers that deal with multiple chromosomes [11–12].

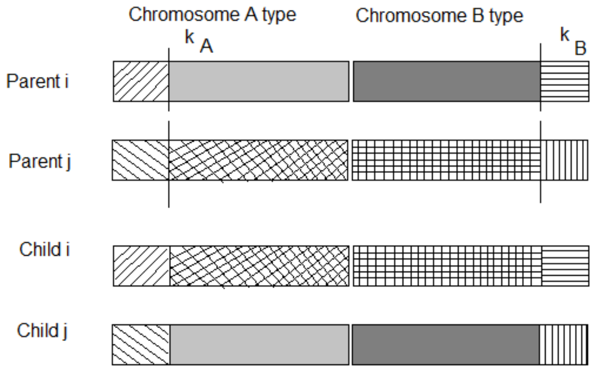

Fig. 8. Crossover in GA-MSC.

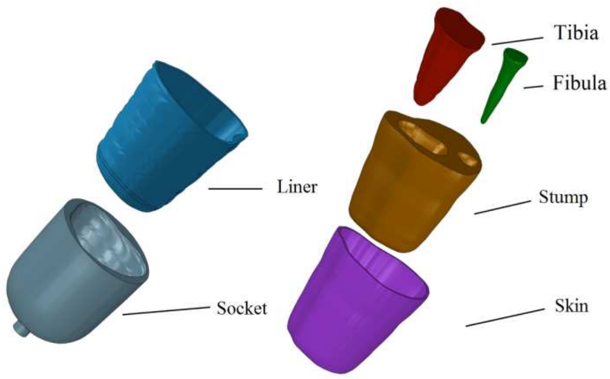

Fig. 9. The parts of modeled assembly stump-liner-socket.

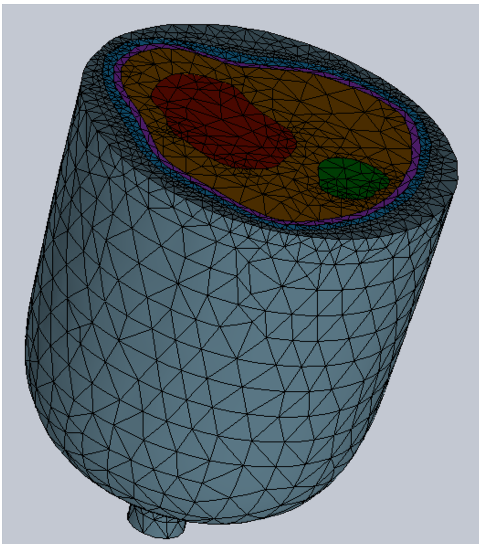

Fig. 10. 3D model of assembly stump-liner-socket meshed.

The population is made by individuals and each individual has two distinct chromosomes of same fixed length. At each generation, selection is made for individual based on tournament method, and crossover and mutation operation are made separately on each chromosome, Fig. 8.

Each generation has two similar operations that operate at each set of chromosomes. There are many crossover operators, each of them both having merits and disadvantages: Single Point Crossover, Two Point Crossover, Intermediate Crossover, Arithmetic Crossover, uniform crossover, elitist crossover and so one [13]. Depending on the problem, this heuristic search can have a better solution in comparison with other methods or not.

We chose the single point crossover algorithm. Each gene from chromosome is binary coded and it corresponds for k_i for chromosome of A type and δ_i for chromosome of B type. Each allele is '0' or '1' and one simple restart mechanism with new random population is made to overcome the saturation algorithm for local minima. Premature convergence is avoided by generating initially an enough diversity as individuals and immigration method for saturation in a point that has a tolerance greater that a prescribed value.

The fitness objective is made by two objectives that apparently are in opposition: the minimum error of model and minimum of elasto-elements. It is known that the precision of algorithm increase with the number of elasto-elements, but if the number of elasto-elements are too high usually more than few hundred, the contribution of small elements to general total is diminished and the results are not satisfactory.

5. EXPERIMENTAL RESULTS OF SIMULATIONS

Data were collected from a lot of two transtibial amputees, each one having a set of 32 repetition of cycle (8 cycles for normal walking, 8 cycles for up the stairs, 8 cycles for down the stairs and 8 cycles normal walking with no rotations). The walking cycle including rotation, walking on difficult terrain and up/down the stairs are the subject of the future research. The stump is MRI scanned and NURB curves are used to approximate the 3D shapes.

Detection contours algorithms are used to separate the various shapes involved in prosthesis: bones, skin, socket, liner. Manual corrections were one object is inserted in another in order to avoid incompatible mesh in FEA analysis. The bodies are inserted one into another taking into account the friction coefficients between surfaces that are put in contact. The bones and soft tissue are modeled as bonded bodies and skin and soft tissue are modeled as tied bodies (friction coefficient 1.0). The coefficient of friction between skin and liner is set to a value between 0.5−0.7 depending on the material of liner.

The properties of skin are modeled to be the same for all the patients without customization. The coefficient of friction between skin and liner is set to a value between 0.5−0.7 depending on the material of liner.

The reconstruction of the bodies is made in SolidWorks 2012 Premium using cloud of points. The individual parts are showed in Fig. 9. Both assembly of tibia, fibula, soft tissue, skin, liner and socket along with meshed as assembly for FEA analysis is presented in Fig. 10–12.

The systems start conventionally for (0, 0) initial point. The first curve that start from (0,0) and end at hysteresis contour is named virgin curve. If the curve is smooth, the model fit very well, Figs. 14–15.

In order to construct the D-F_t curve a simulation model and FEA stress analysis is used in SolidWorks.

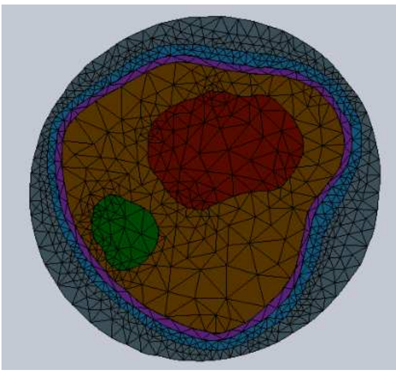

Fig. 11. 3D model of assembly stump-liner-socket meshed, upper view.

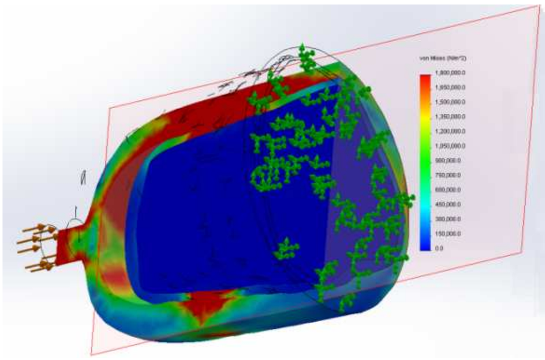

Fig. 12. 3D model of assembly stump-liner-socket meshed.

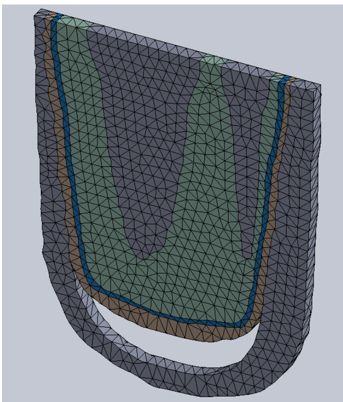

Fig. 13. Slice extracted from FEM 3D model.

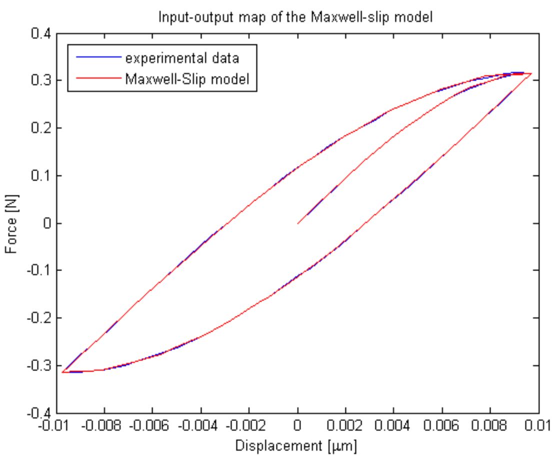

Fig. 14. Maxwell-Slip model, M = 10 elements, piezzo-electric actuator, GA-MSC algorithm.

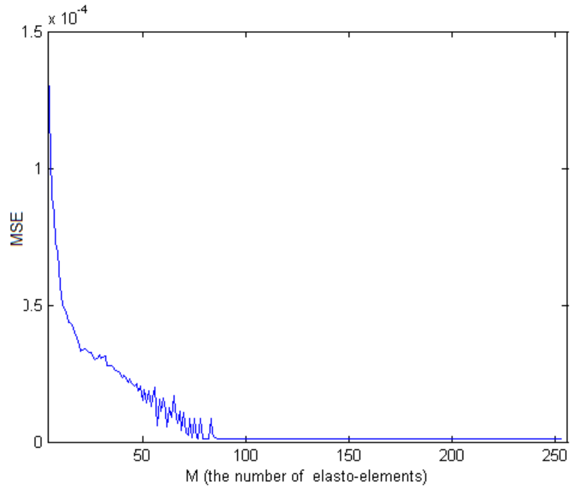

Fig. 15. MSE evolution, Maxwell-Slip model, M = 1–256 elements, piezzo-electric actuator, GA-MSC algorithm, 70 samples.

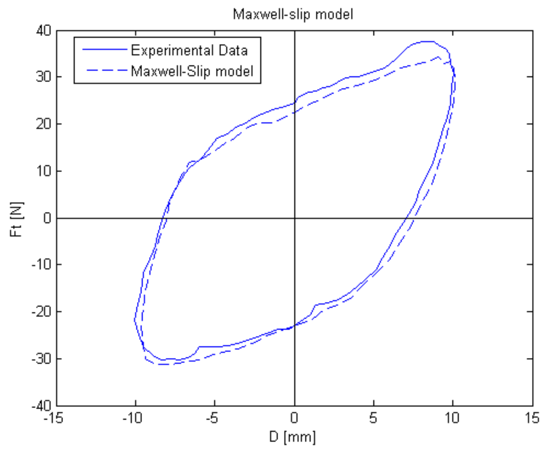

Fig. 16. Maxwell-Slip model, M = 80 elements.

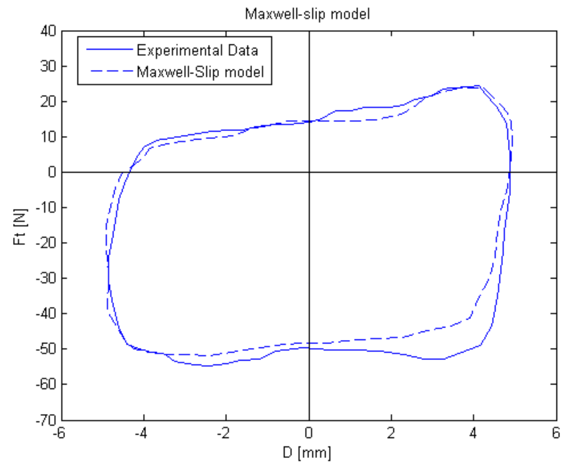

Fig. 17. Maxwell-Slip model, M = 140 elements.

We remark that is not practically possible to extract the displacement of tetrahedral nodes at the interface level from skin surface, Figs. 10–12. The usage of section from 3D is not feasible because of difficulties given by SolidWorks software that cannot permit the extract to values from nodes from a shell that are locate inside other shells.

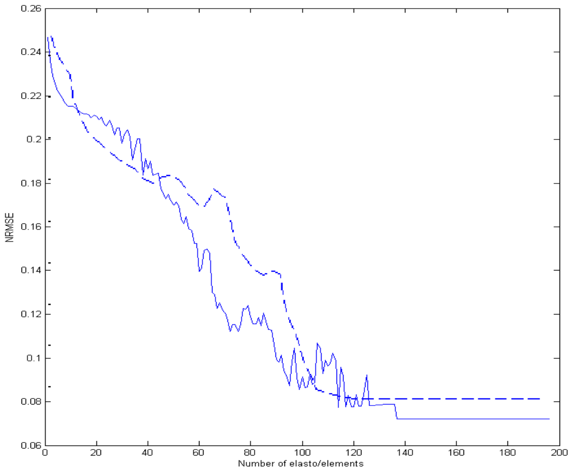

Fig. 18. LR-solid line, GA-MSC algorithm – dashed line.

Our approach is to extract a thin shell (in our case the thickness is set to 5 mm, by empirical trials) and to divide the force that acts on upper, and lower parts according to area covered by this shell in the upper part and lower part of the stump, Fig. 13. The lateral parts are considered fixed boundaries and it are marked in the last shell by socket walls (that are rigid in all the prosthetic applications).

The algorithm for measurement of F_t and displacement is made at each step from start to a discrete moment of time t, and t + 1 the next and so one. The solution that uses our proposed algorithm is compared with the simple one, the linear regression (LR). LR uses preassigned values for Δ_i, each value is given by [7]:

$$\Delta_i = \frac{i}{M} \cdot \max\{|x|\}, \quad i = 1, \ldots M \ , \tag{8}$$

$$F(t) = \sum_{i=1}^{M} k_i \cdot \delta_i(t) + e(t) \ . \tag{9}$$

The presliding force is represented by LR(M, k) model (linear model) given by (9) subject to (3). Minimization of error $e(t)$ leads to a linear estimator for k.

We use Normalized Mean Square Error (NMSE) as overall estimator between experimental values and model values. The fit model is given by (n samples):

$$fit_{NMSE} = 1 - NMSE \ , \tag{10}$$

$$NMSE = \sqrt{\sum_{i=1}^{n} (y_i - \hat{y}_i)^2} \ / \sqrt{\sum_{i=1}^{n} (y_i - \bar{y})^2} \ . \tag{11}$$

The value $\bar{y}$ represents the mean of experimental data and the caret denotes the estimated value. The results showed that for the first hysteresis curve in some situations genetic algorithms give better results in comparison with LR method, Figs. 16–18.

6. CONCLUSIONS

A model for skin hysteresis in prosthetic application is proposed in this paper. For some hysteresis curves, the method can reduce the error and as sequel to improve the fitness of model.

The results are very encouraging, but we can see from these preliminary results that in order to obtain a fit model, the experimental data must be preprocessed in order to obtain a smoother curve.

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