Analysis of muscles behaviour.
Part I. The computational model of muscle

WIKTORIA WOJNICZ*, EDMUND WITTBRODT*

Mechanics and Strength of Materials Department, Mechanical Engineering Faculty,
Gdańsk University of Technology, Gdańsk, Poland.

The purpose of this paper is to present the computational model of skeletal muscle, which was treated as the structure of different mechanical properties. The method of identification of those properties is described in detail. In addition, the method of quantitative and qualitative verification of this model is proposed. Applying this computational model, the forces of real muscles can be evaluated without using any optimization technique. Such an approach will be described in the part II of the paper.

Key words: *muscle, modelling, identification, simulation, marker*

1. Introduction

Nowadays the mathematical modelling is more and more frequently used in research of the behaviour of biological systems, especially in the process of rehabilitation, in medicine and sports biomechanics. The reason for this is the need for identification and connection of the series of phenomena occurring in the living form. In the majority of cases, the direct measurements of data that describe those phenomena cause disturbances of organism's functions or permanent damage to its organs. For that reason the mathematical modelling has been used during the identification of cause-and-effect relationships. It consists in creating the computational model of the phenomenon examined, identifying its parameters and performing the computer simulation of its behaviour. The correctness of the computational model proposed results from the verification, i.e. the comparison between numerical simulation outcomes and the data measured by the methods that do not affect the function of organism.

From the mechanical point of view the behaviour of striated skeletal muscle (being further named the muscle) is frequently modelled by using the Hill-type muscle models (van der BOGERT et al. [1], CAMILLERI and HULL [2], GARNER and PANDY [6], LEMOS et al. [10], MEIJER et al. [14]). Due to this approach the phenomenon of contraction can be described on a macroscopic or microscopic level. On a macroscopic level a real muscle is treated as the whole and its computational model is composed of several different rheological elements (predominantly of elastic, viscoelastic and forcible character) that describe mechanical properties of muscle based on the following parameters: the peak isometric force, the optimal muscle-fiber length, the fiber pennation angle, the tendon slack length and the maximum intrinsic shortening velocity of muscle (DELP [4], GARNER and PANDY [6]). The values of the parameters are evaluated in accordance with given assumptions by using the data obtained from cadaver measurements. Moreover, this way of modelling requires both static and dynamic characteristics of each real muscle being modelled. Both characteristics cannot be determined

based on the muscles belonging to the real muscles' group and that is why they are inputted as analytical functions (LEMOS et al. [10], CAMILLERI and HULL [2], REHBINDER and MARTIN [17]).

Within the last few years, using the finite element method, there were created models describing the behaviour of muscle on a microscopic level (van DONKELLAR et al. [5], JOHANSSON et al. [9], LEMOS et al. [10], YUCESOY et al. [22]). In this method, the muscle is treated as the area of discretized continuum. In this approach, we have to evaluate a large number of material constants that cannot be directly determined in the alive muscle having anisotropic properties. Moreover, static and dynamic characteristics of isolated muscle are used for the description of finite element viscoelastic properties. The adjustment of both characteristics that are on a macroscopic level to a microscopic level is still a troublesome problem.

Considering the above mentioned problems, we can see that there is still the demand for computational models of muscle that permit us to reflect the truthful mechanical properties of alive muscle and the methods of identifying its parameters.

The main goal of this paper is to present: 1) a new approach to modelling the muscle behaviour, 2) a comparatively simple method for identification of muscle's computational model parameters, which can be used for alive muscles; 3) the method of quantitative and qualitative verification of the muscle model proposed.

2. The computational model of muscle

Based on the physiology of a real fusiform muscle (figure (A)), in which serially linked parts can be distinguished, the computational model of muscle was created. These parts have different viscoelastic properties, because they contain dissimilar quantities of tissue fluids. In figure (B), the rheological muscle model is presented. It is composed of two passive parts (modelling the action of tendon insertion I and tendon origin V) and three active parts (modeling three contractile parts of belly – II, III, IV) (WITTBRODT and WOJNICZ [18], WOJNICZ and WITTBRODT [19], WOJNICZ [21]). Their elastic properties are represented by the stiffness coefficients $K = \{K_0, K_1, K_2, K_3, K_4\}$, and viscous properties – by the damping coefficients $L = \{L_0, L_1, L_2, L_3, L_4\}$, thus reduced masses of these parts are expressed by the mass coefficients $m = \{m_0, m_1, m_2, m_3, m_4\}$. In the active parts of muscle's belly (II, III, IV), there were placed force ele-

ments generating two internal forces of opposite directions: $P_{11}^w = P_{11}^w(t)$ and $-P_{12}^w = -P_{12}^w(t)$; $P_{21}^w = P_{21}^w(t)$ and $-P_{22}^w = -P_{22}^w(t)$; $P_{31}^w = P_{31}^w(t)$ and $-P_{32}^w = -P_{32}^w(t)$. The displacements of reduced muscle mass elements $x = x(t) \in \{x_0(t), x_1(t), x_2(t), x_3(t), x_4(t)\}$ reflect the displacements of the points that are the boundaries between the chosen parts of a real muscle. The junction of tendon origin with the non-movable basis models the fixing of a real muscle to an immovable bone. The external force $P_{ext} = P_{ext}(t)$ can be applied to the tendon insertion I (which is fixed to the moving bone). This force is always directed to the elongation of muscle and never causes its active contraction (because only force elements can cause this behaviour).
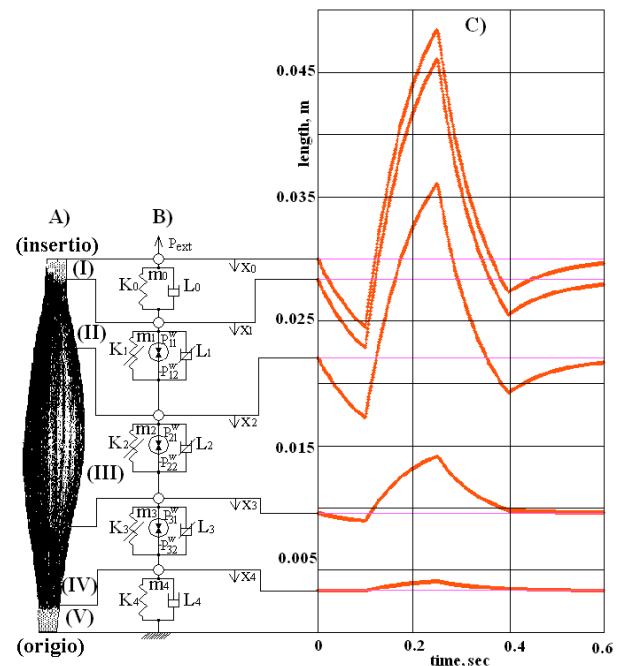
The fusiform muscle: A) the real form, B) its rheological model, C) the example of simulation of behaviour of muscle model, for which the following data were used:
$K = \{31000, 5000, 1500, 5000, 31000\}$ N/m,
$L = \{3100, 500, 150, 500, 3100\}$ N·s/m,
$m = \{0.00005, 0.01187, 0.076, 0.01187, 0.00011\}$ kg,
$P_{ext} = 30 \cdot \sigma(t - 0.1) - 30 \cdot \sigma(t - 0.25)$ N,
$P_{11}^w = P_{12}^w = P_{31}^w = P_{32}^w = 5 \cdot \sigma(t - 0.00001) - 5 \cdot \sigma(t - 0.4)$ N,
$P_{21}^w = P_{22}^w = 10 \cdot \sigma(t - 0.00001) - 10 \cdot \sigma(t - 0.4)$ N

It should be noted that the proposed computational model of muscle ensures different forces at the tendon origin V and the tendon insertion I which is in agreement with the experimental data reported (MASS et al. [12], [13]).

If the behaviour of muscle model is consistent with the physiological behaviour of the alive muscle, we can assume the restrictions of the displacements x, their

velocities $\dot{x} = \dot{x}(t) \in \{\dot{x}_0(t), \dot{x}_1(t), \dot{x}_2(t), \dot{x}_3(t), \dot{x}_4(t)\}$, the force $F_{ins} = F_{ins}(t)$ acting in the junction of the tendon insertion with the belly and the force $F_{or} = F_{or}(t)$ acting in the junction of the tendon origin with the belly:

$$-F_{ins} = L_0 \cdot (\dot{x}_0 - \dot{x}_1) + K_0 \cdot (x_0 - x_1), \qquad (1)$$

$$F_{or} = L_3 \cdot (\dot{x}_4 - \dot{x}_3) + K_3 \cdot (x_4 - x_3) + P_{32}^w. \qquad (2)$$

During numerical simulations the confirmation of the state of muscle model (i.e., whether this model is in the "admissible state") results from the examination of fulfilment of the above-mentioned restrictions.

3. The identification of parameters

The identification of the parameters of muscle model is based on the evaluation of its stiffness K and damping coefficients L. It has been assumed that the viscoelastic properties of non-excited muscle model are the same as those of the excited muscle model. Moreover, these properties are constant during each little lengthening/shortening of real muscle under consideration.

The mass coefficients m can be estimated on the basis of muscle geometric and mass dimensions by using computed tomography or magnetic resonance imaging or USG technique (NARICI [16], DANIEL et al. [3]). After that three-dimensional images of muscle and the data of dimensions/mass/volume/density of its parts might be obtained. These data are the basis for: 1) the virtual dividing of this muscle up into parts; 2) the evaluation of their masses m; 3) the determination of points indicating boundaries between chosen parts.

The coefficients representing viscoelastic properties must be evaluated within the range of the length of real muscle belonging to the muscles' group under examination and the muscle must be kept inactive. In an initial state of identification, the length of the muscle fixed to the bone reaches its minimum, while in a final state, this length reaches its maximum. Then the input data i.e. identified parameters, are as follows: the displacements of chosen points lying on the real muscle x, their velocities $\dot{x}$, their accelerations $\ddot{x} = \ddot{x}(t) \in \{\ddot{x}_0(t), \ddot{x}_1(t), \ddot{x}_2(t), \ddot{x}_3(t), \ddot{x}_4(t)\}$ and the external lengthening force P_{ext}.

The process of identification is carried out in two stages. During the first stage the muscle is subjected to stepwise lengthening several times from the initial state to the final state due to the action of the external force P_{ext}, which is the step increasing function. During the

second stage this muscle is subjected to stepwise shortening several times from the final state to the initial state caused by the external force P_{ext}, which is the step decreasing function. Based on the single-step lengthening/shortening of the muscle, the stiffness and damping coefficients are evaluated. The number of stepwise lengthening/shortening of muscle, occurring several times, influence the accuracy of evaluating its viscoelastic properties depending on the length and the kind of work performed by the muscle parts. It is worth noticing that this accuracy also depends on the precision of measurements of kinematical and forcible data.

The single step i-th lengthening/shortening of muscle is held at the time interval $[t_{i–1}; t_i]$. During this interval the length of muscle is being changed from $l_{i–1}$ to l_i due to the action of the external force P_{ext} influencing the tendon insertion. The value of this force is being increased/decreased stepwise from $P_{ext}(t_{i–1})$ to $P_{ext}(t_i)$.

At the boundary times $t_{i–1}$ and t_i, the muscle is in a steady state: the displacements of chosen points are constant, their velocities and accelerations equal zero, i.e. $(x_k(t_{i–1}) = C_k^{i–1}) \wedge (x_k(t_i) = C_k^i) \wedge (\dot{x}_k(t_{i–1}) = 0) \wedge (\ddot{x}_k(t_{i–1}) = 0) \wedge (\ddot{x}_k(t_i) = 0)$ for $k = 0, 1, 2, 3, 4$, where $(C_k^{i–1} = \text{const}) \wedge (C_k^i = \text{const})$. At the time t_i, one can evaluate the stiffness coefficients (K_0^p, K_1^p, K_2^p, K_3^p, K_4^p) that describe elastic properties of this muscle in the defined range of its length $(l_{i–1}, l_i]$, by using the following system of equations:

$$
\begin{aligned}
K_0^p &= \frac{-P_{ext}(t_i)}{x_0(t_i) - x_1(t_i)}, \\[2mm]
K_1^p &= \frac{K_0^p \cdot (x_0(t_i) - x_1(t_i))}{x_1(t_i) - x_2(t_i)}, \\[2mm]
K_2^p &= \frac{K_1^p \cdot (x_1(t_i) - x_2(t_i))}{x_2(t_i) - x_3(t_i)}, \qquad (3) \\[2mm]
K_3^p &= \frac{K_2^p \cdot (x_2(t_i) - x_3(t_i))}{x_3(t_i) - x_4(t_i)}, \\[2mm]
K_4^p &= \frac{K_3^p \cdot (x_3(t_i) - x_4(t_i))}{x_4(t_i)},
\end{aligned}
$$

where:

K_k^p ($k = 0, 1, 2, 3, 4$) – the stiffness coefficient,

p – the kind of work performed by muscle (the stiffness coefficients K_k^p for $p = w$ are the coefficients during the lengthening of muscle, thus for $p = s$ during the shortening of muscle).

For the purpose of evaluating damping coefficients at the time interval $[t_{i–1}, t_i]$ the kinematical

and forcible data must be defined at the time $t_m \in (t_{i-1}; t_i)$. It is worth noticing that damping coefficients can be evaluated when the velocities of chosen points located on the real muscle are different from zero. This situation can exclusively happen during the dynamic process. Then at the time t_m, the damping coefficients (L_0^p, L_1^p, L_2^p, L_3^p, L_4^p) that describe viscous properties of muscle in the defined range of its length (l_{i-1}, l_i] can be evaluated from the following system of equations:

$$L_0^p = \frac{-P_{\text{ext}}(t_m) - K_0^p \cdot (x_0(t_m) - x_1(t_m)) - m_0 \cdot \ddot{x}_0(t_m)}{\dot{x}_0(t_m) - \dot{x}_1(t_m)},$$

$$L_1^p = \frac{K_0^p \cdot (x_0(t_m) - x_1(t_m)) - K_1^p \cdot (x_1(t_m) - x_2(t_m)) + L_0^p \cdot (\dot{x}_0(t_m) - \dot{x}_1(t_m)) - m_1 \cdot \ddot{x}_1(t_m)}{\dot{x}_1(t_m) - \dot{x}_2(t_m)},$$

$$L_2^p = \frac{K_1^p \cdot (x_1(t_m) - x_2(t_m)) - K_2^p \cdot (x_2(t_m) - x_3(t_m)) + L_1^p \cdot (\dot{x}_1(t_m) - \dot{x}_2(t_m)) - m_2 \cdot \ddot{x}_2(t_m)}{\dot{x}_2(t_m) - \dot{x}_3(t_m)}, \quad (4)$$

$$L_3^p = \frac{K_2^p \cdot (x_2(t_m) - x_3(t_m)) - K_3^p \cdot (x_3(t_m) - x_4(t_m)) + L_2^p \cdot (\dot{x}_2(t_m) - \dot{x}_3(t_m)) - m_3 \cdot \ddot{x}_3(t_m)}{\dot{x}_3(t_m) - \dot{x}_4(t_m)},$$

$$L_4^p = \frac{K_3^p \cdot (x_3(t_m) - x_4(t_m)) - K_4^p \cdot x_4(t_m) + L_3^p \cdot (\dot{x}_3(t_m) - \dot{x}_4(t_m)) - m_4 \cdot \ddot{x}_4(t_m)}{\dot{x}_4(t_m)},$$

where:

L_k^p ($k = 0, 1, 2, 3, 4$) – the damping coefficient,

p – the kind of work performed by muscle (the damping coefficients L_k^p for $p = w$ are the coefficients during the lengthening of muscle, thus for $p = s$ during the shortening of muscle).

Taking into account physiological alterations in the real muscle (the fatigue, the change of its properties, e.g., as a result of relaxation of force/length) its mechanical properties can be changed. Therefore the parameters of muscle model should be defined under firmly fixed conditions (the proper temperature, moisture and pressure) in a possibly short time period.

4. The identification of muscle model internal forces

The internal forces in the muscle model can be identified, provided that its parameters have been evaluated and the following time-dependent experi-

mental data have been obtained: the displacements x of chosen points placed on the real muscle, their velocities $\dot{x}$, their accelerations $\ddot{x}$ and the value of the external force P_{ext}.

At the initial stage of this identification, the internal forces are defined by using the model of muscle generating balanced forces. If for this model these internal forces cannot be uniquely evaluated, they must be defined by using the model of muscle generating unbalanced forces.

4.1. The model of muscle generating balanced forces

The model of muscle, in which each force element generates two internal forces having the same magnitude but the opposite direction, is named the model of muscle generating balanced forces, i.e. $[P_1^w(t) = P_1^w = P_{11}^w = P_{12}^w] \wedge [P_2^w(t) = P_2^w = P_{21}^w = P_{22}^w] \wedge [P_3^w(t) = P_3^w = P_{31}^w = P_{32}^w]$. This model might be used if the unique solution of the following system of equations exists:

$$P_1^w = m_1 \cdot \ddot{x}_1 + L_0 \cdot (\dot{x}_1 - \dot{x}_0) + K_0 \cdot (x_1 - x_0)$$
$$+ L_1 \cdot (\dot{x}_1 - \dot{x}_2) + K_1 \cdot (x_1 - x_2),$$

$$P_2^w - P_1^w = m_2 \cdot \ddot{x}_2 + L_1 \cdot (\dot{x}_2 - \dot{x}_1) + K_1 \cdot (x_2 - x_1)$$
$$+ L_2 \cdot (\dot{x}_2 - \dot{x}_3) + K_2 \cdot (x_2 - x_3),$$

$$P_3^w - P_2^w = m_3 \cdot \ddot{x}_3 + L_2 \cdot (\dot{x}_3 - \dot{x}_2) + K_2 \cdot (x_3 - x_2)$$
$$+ L_3 \cdot (\dot{x}_3 - \dot{x}_4) + K_3 \cdot (x_3 - x_4), \quad (5)$$

$$P_3^w = -m_4 \cdot \ddot{x}_4 - L_3 \cdot (\dot{x}_4 - \dot{x}_3)$$
$$- K_3 \cdot (x_4 - x_3) - L_4 \cdot \dot{x}_4 - K_4 \cdot x_4.$$

4.2. The model of muscle generating unbalanced forces

In this model, each force element generates non-equal internal forces (P_{11}^w, P_{12}^w, P_{21}^w, P_{22}^w, P_{31}^w, P_{32}^w). Its rheological representation is shown in figure (B), and its mathematical form is described by the following system of five differential equations:

$$m_0 \cdot \ddot{x}_0 + L_0 \cdot (\dot{x}_0 - \dot{x}_1)$$
$$+ K_0 \cdot (x_0 - x_1) = -P_{\text{ext}},$$

$$m_1 \cdot \ddot{x}_1 + L_0 \cdot (\dot{x}_1 - \dot{x}_0) + K_0 \cdot (x_1 - x_0)$$
$$+ L_1 \cdot (\dot{x}_1 - \dot{x}_2) + K_1 \cdot (x_1 - x_2) = P_{11}^w,$$

$$m_2 \cdot \ddot{x}_2 + L_1 \cdot (\dot{x}_2 - \dot{x}_1) + K_1 \cdot (x_2 - x_1)$$
$$+ L_2 \cdot (\dot{x}_2 - \dot{x}_3) + K_2 \cdot (x_2 - x_3) = P_{21}^w - P_{12}^w, \quad (6)$$

$$m_3 \cdot \ddot{x}_3 + L_2 \cdot (\dot{x}_3 - \dot{x}_2) + K_2 \cdot (x_3 - x_2)$$
$$+ L_3 \cdot (\dot{x}_3 - \dot{x}_4) + K_3 \cdot (x_3 - x_4) = P_{31}^w - P_{22}^w,$$

$$m_4 \cdot \ddot{x}_4 + L_3 \cdot (\dot{x}_4 - \dot{x}_3) + K_3 \cdot (x_4 - x_3)$$
$$+ L_4 \cdot \dot{x}_4 + K_4 \cdot x_4 = -P_{32}^w.$$

For the purpose of a unique evaluation of the system of equations (6) it has been assumed that in the force elements of muscle's model part *III*, the same internal forces are generated $P_{21}^w(t) = P_{22}^w(t) = P_2^w(t) = P_2^w$. This assumption is based on the physiology of a fusiform muscle: the number of contractile fibers in the high and low sections of the middle part of the real belly are approximately the same. Aiming at determining five unknown internal forces $P_{11}^w(t)$, $P_{12}^w(t)$, $P_2^w(t)$, $P_{31}^w(t)$, $P_{32}^w(t)$ generated at the time *t*, the following system of five equations must be solved:

$$P_a(t) = P_p(t) - P_{\text{ext}}(t),$$

$$P_{11}^w = m_1 \cdot \ddot{x}_1 + L_0 \cdot (\dot{x}_1 - \dot{x}_0) + K_0 \cdot (x_1 - x_0)$$
$$+ L_1 \cdot (\dot{x}_1 - \dot{x}_2) + K_1 \cdot (x_1 - x_2),$$

$$P_2^w - P_{12}^w = m_2 \cdot \ddot{x}_2 + L_1 \cdot (\dot{x}_2 - \dot{x}_1) + K_1 \cdot (x_2 - x_1)$$
$$+ L_2 \cdot (\dot{x}_2 - \dot{x}_3) + K_2 \cdot (x_2 - x_3), \quad (7)$$

$$P_{31}^w - P_2^w = m_3 \cdot \ddot{x}_3 + L_2 \cdot (\dot{x}_3 - \dot{x}_2) + K_2 \cdot (x_3 - x_2)$$
$$+ L_3 \cdot (\dot{x}_3 - \dot{x}_4) + K_3 \cdot (x_3 - x_4),$$

$$P_{32}^w = -m_4 \cdot \ddot{x}_4 - L_3 \cdot (\dot{x}_4 - \dot{x}_3)$$
$$- K_3 \cdot (x_4 - x_3) - L_4 \cdot \dot{x}_4 - K_4 \cdot x_4,$$

where:

$$P_p(t) = A_2 \cdot \ddot{x}_0 + A_1 \cdot \dot{x}_0 + A_0 \cdot x_0$$
$$+ \int_0^t \left(\sum_{i=0}^7 B_i \cdot e^{s_i \cdot (t-\tau)} \right) \cdot x_0(\tau) \, d\tau, \quad (8)$$

$$P_a(t) = \int_0^t \left(\sum_{i=0}^7 D_{i1} \cdot e^{s_i \cdot (t-\tau)} \right) \cdot P_{11}^w(\tau) \, d\tau$$
$$+ \int_0^t \left(\sum_{i=0}^7 D_{i2} \cdot e^{s_i \cdot (t-\tau)} \right) \cdot P_{12}^w(\tau) \, d\tau$$
$$+ \int_0^t \left(\sum_{i=0}^7 D_{i3} \cdot e^{s_i \cdot (t-\tau)} \right) \cdot P_2^w(\tau) \, d\tau$$
$$+ \int_0^t \left(\sum_{i=0}^7 D_{i4} \cdot e^{s_i \cdot (t-\tau)} \right) \cdot P_{31}^w(\tau) \, d\tau$$
$$+ \int_0^t \left(\sum_{i=0}^7 D_{i5} \cdot e^{s_i \cdot (t-\tau)} \right) \cdot P_{32}^w(\tau) \, d\tau, \quad (9)$$

$P_p(t)$ – the passive component that is evaluated at whichever time *t* by using measured kinematical data of muscle insertion (its displacement x_0, velocity $\dot{x}_0$ and acceleration $\ddot{x}_0$),

$P_a(t)$ – the active component dependent on five calculated internal forces $P_{11}^w(t)$, $P_{12}^w(t)$, $P_2^w(t)$, $P_{31}^w(t)$, $P_{32}^w(t)$,

A_0, A_1, A_2, B_0, B_i, s_i, D_{i1}, D_{i2}, D_{i3}, D_{i4}, D_{i5} (i = 0, 1, 2, 3, 4, 5, 6, 7) – the coefficients dependent on the mass coefficients *m*, stiffness coefficients *K*, damping coefficients *L* and the kind of work of muscle parts (lengthening or shortening).

5. The method of verification and the example simulation

The proposed model of muscle allows the quantitative and qualitative verification to be carried out. The quantitative verification consists in comparing the following data measured and calculated in computer simulation: 1) the displacements *x* of points placed on the muscle surface, their velocities $\dot{x}$ and accelerations $\ddot{x}$, 2) the forces in the tendon insertion $P_{\text{ten\_in}}(t)$ and the tendon origin $P_{\text{ten\_or}}(t)$:

$$P_{\text{ten\_in}}(t) = P_{\text{ext}}(t), \quad (10)$$

$$P_{ten\_or}(t) = -L_4 \cdot \dot{x}_4 - K_4 \cdot x_4. \qquad (11)$$

In order to register timing displacements of chosen points one can use markers (e.g. fluorescent polystyrene spheres or sonographic crystals) glued to the muscle's surface (van DONKELAAR et al. [5], HUIJING [8]). Also one can use markers implanted into the inwards of muscle (LEMOS et al. [10]), but this method is invasive and might disturb the functioning of living organisms. Using the dynamometer hitched onto the tendon insertion one can measure the true value of the external force P_{ext} (HERZOG [7]).

The qualitative verification consists in comparing the EMG-signals [de LUCA [11]) measured by invasive or surface electrodes placed on the defined parts of examined alive muscle belly with calculated internal forces (see the chapter/part 4).

In order to prove that the muscle model proposed is "working", the example of computer simulation outcomes are presented in figure (C). The outcomes result in forward dynamic task solving.

6. Conclusions

Treating a real muscle as the structure of different mass-elastic-viscous properties, which can be contracted under the action of internal forces, the computational model of muscle was created. During the identification of muscle parameters the viscoelastic properties of its model can be evaluated based on experimental data referred to timing displacements of markers (attached to the surface of muscle under consideration) and the force measured by dynamometer (hitched onto its tendon insertion). Furthermore, reflecting the contractility of muscle fibers, the internal forces generated in force elements of muscle's model can be evaluated during the identification of internal forces.

This computational model can be used for solving the forward and inverse dynamic tasks. Restricting some values of the muscle model (the models have to be in "admissible states"), it is possible to represent the working of neural system, which, by means of sensory organs (muscle spindles and Golgi tendon organs), keeps the path of the behaviour of all muscles and tries not to allow their damage.

In our new way of modelling, we can neglect the series of data describing the parameters of a Hill-type muscle model that do not reflect the behaviour of muscles belonging to a real muscles' group. Moreover, the proposed method of identification permits us to simply obtain the parameter necessary for performing the computer simulation.

Using the muscle model proposed, the computational model of muscles' group affecting a defined human joint might be simply created. Therefore the participation of real muscles in the movement of the joint examined can be evaluated without using any optimization technique (this method is described by WOJNICZ and WITTBRODT [20] in part II of their paper).

The numerical simulations helpful in developing the method presented in this paper were performed using the computers owned by "Centrum Informatyczne Trójmiejskiej Akademickiej Sieci Komputerowej" in Gdańsk, Poland.

References

[1] van den BOGERT A.J., GERRITSEN K.G.M., COLE G.K., *Human muscle modelling from a user's perspective*, Journal of Electromyography and Kinesiology, 1998, 8, 119–124.

[2] CAMILLERI M.J., HULL M.L., *Are the maximum shortening velocity and the shape parameters in a Hill-type model of whole muscle related to activation?* Journal of Biomechanics, 2005, 38, 2172–2180.

[3] DANIEL M., IGLIČ A., KRALJ-IGLIČ V., KONVIČKOVÁ S., *Computer system for definition of the quantitative geometry of musculature from CT images*, Computer Methods in Biomechanics and Biomedical Engineering, 2005, 8, 25–29.

[4] DELP S.L., *Surgery simulation: a computer graphics system to analyze and design musculoskeletal reconstructions of the lower limb*, PhD thesis, 1990, Stanford University Press, Stanford.

[5] van DONKELAAR C.C., WILLEMS P.J.B., MUIJTJENS A.M.M., DROST M.R., *Skeletal muscle transverse strain during isometric contraction at different lengths*, Journal of Biomechanics, 1999, 32, 755–762.

[6] GARNER B.A., PANDY M.G., *Estimation of musculotendon properties in the human upper limb*, Annals of Biomedical Engineering, 2003, 31, 207–220.

[7] HERZOG W., *Muscle synergies during voluntary movement*, Proceedings of XI International Biomechanics Seminar, 1998, Wrocław, Poland, 7–22.

[8] HUIJING P.A., *Parameter interdependence and success of skeletal muscle modelling*, Human Movement Science, 1995, 14, 443–486.

[9] JOHANSSON T., MEIER P., BLICKHAN R., *A finite element model for the mechanical analysis of skeletal muscles*, Journal of Theoretical Biology, 2000, 206, 131–149.

[10] LEMOS R.R., EPSTEIN M., HERZOG W., WYVILL B., *A framework for structured modelling of skeletal muscle*, Computer Methods in Biomechanics and Biomedical Engineering, 2004, 7, 305–317.

[11] de LUCA C.J., *The use of surface electromyography in biomechanics*, Journal of Applied Biomechanics, 1997, 13, 135–163.

[12] MAAS H., BAAN G.S., HUIJING P.A., *Muscle force is determined also by muscle relative position: isolated effects*, Journal of Biomechanics, 2004, 37, 99–110.

[13] MAAS H., LENTI T.M., TIIHONEN V., KOMULAINEN J., HUIJING P.A., *Controlled intermittent shortening contractions of a muscle–tendon complex: muscle fibre damage and effects on force transmission from a single head of rat EDL*, Journal of Muscle Research and Cell Motility, 2005, 26, 259–273.

[14] MEIJER K., GROOTENBOER H.J., KOOPMAN H.F.J.M., van der LINDEN B.J.J.J., HUIJING P.A., *A Hill type model of rat medial gastrocnemius muscle that accounts for shortening history effects*, Journal of Biomechanics, 1998, 31, 555–563.

[15] MROZOWSKI J., AWREJCEWICZ J., *Podstawy biomechaniki*, Politechnika Łódzka, Łódź, 2004.

[16] NARICI M., *Human skeletal muscle architecture studied in vivo by non-invasive imaging techniques: functional significance and applications*, Journal of Electromyography and Kinesiology, 1999, 9, 97–103.

[17] REHBINDER H., MARTIN C., *A control theoretic model of the forearm*, Journal of Biomechanics, 2001, 34, 741–748.

[18] WITTBRODT E., WOJNICZ W., *Modelling and simulation of striated skeletal muscle and groups of muscles in arm–forearm system 2D*, Proceedings of the International Conference *Biomechanics 2003*, Acta of Bioengineering and Biomechanics, 2003, Vol. 5, Supplement 1, 551–556.

[19] WOJNICZ W., WITTBRODT E., *The modelling of behaviour of skeletal muscles in a dynamic state* (in Polish), Proceedings of the International Conference *Biomechanics 2004*, Acta of Bioengineering and Biomechanics, 2004, Vol. 6, Supplement 1, 404–408.

[20] WOJNICZ W., WITTBRODT E., *Analysis of muscles behaviour*. Part II. *The computational model of muscles' group acting on the elbow joint*, Acta of Bioengineering and Biomechanics, 2010 (to be published).

[21] WOJNICZ W., *Modelowanie i symulacja zachowania zespołu mięśni szkieletowych układu ramię–przedramię*, PhD Thesis, 2009, Technical University of Łódź, Poland.

[22] YUCESOY C.A., KOOPMAN B.H.F.J.M., HUIJING P.A., GROOTENBOER H.J., *Three dimensional finite element modeling of skeletal muscle using a two-domain approach: linked fiber-matrix mesh model*, Journal of Biomechanics, 2002, 35, 1253–1262.