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Original Paper

The necessity of physiological muscle parameters for computing the muscle forces: application to lower extremity loading during pedalling

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The aim of this study is to determine how the use of physiological parameters of muscles is important. This work is focused on musculoskeletal loading analysis during pedalling adopting two approaches: without (1) and with (2) the use of physiological parameters of muscles. The static-optimization approach together with the inverse dynamics problem makes it possible to obtain forces in individual muscles of the lower extremity. Input kinematics variables were examined in a cycling experiment. The significant difference in the resultant forces in one-joint and two-joint muscles using the two different approaches was observed.

Key words: *muscle forces, static optimization, Hill type muscle model, pedalling, physiological parameters of muscles*

1. Introduction

The aim of this work was to demonstrate whether the physiological parameters of muscles are necessary for computing the muscle forces. Since the determination of physiological muscle parameters is difficult, these data in the literature differ (see, for example, [1] and [2]) and the sensitivity of result to the input parameters is crucial (see [3] and [4]), it would be easier to determine the muscle forces without the use of these uncertain parameters. Computation without these parameters, i.e. only with the use of joint moments and other external forces, would be much quicker, easier and less time-consuming.

The problems posed by a human model with muscles are underdetermined, thus the distribution problem and static optimization approach are used. The advantage of this method lies in its computational efficiency. Even though the dynamic optimization can provide more realistic estimates of muscle forces, it is much more computationally expensive; therefore it would be necessary to use a more simple model to keep the computation time in reasonable limits. The differences between the static and dynamic optimization were described, for example, by ANDERSON [5].

To create our model, we did not use the human body software developed by DELP [1], but as others did [5], [6], we created our own model that fulfilled our requirements more precisely. Since it is necessary to know the kinematics variables of the movement and the force acting on the lower extremity, they are either measured in a cycling experiment [6] as we did, or the EMG-driven model is used [7] together with the inverse dynamics.

The static optimization procedure needs a cost function to be minimized. There are several ways to do it [8]. One of the approaches is to minimize the sum of the muscle forces divided by the cross-sectional area of the muscle to the power of two (or three, or even higher power) [3]. Other approach is to minimize only the sum of muscle forces to the power

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of two (or three, or even higher power). Finally, the cost function can be set as the sum of the muscle activation to the power of two [9]. As we did two different computations, we used two last approaches mentioned above.

First of all, we computed the muscle forces without knowing the physiological parameters of the muscles of the lower limb. Only the Newton–Euler's equations were used. In this case, the cost function was the sum of muscle forces to the power of two. The second case was the computation using the Newton–Euler's equations together with the equation of muscle dynamics and the consideration of the muscle activation and physiological parameters of muscles. In this case, the sum of the muscle activation to the power of two was used.

Furthermore, we investigated the suitability of several optimization approaches (Matlab© Optimization Toolbox was used) and the most appropriate one was chosen for the further computations. The results show how the use of physiological parameters of muscles is crucial in the computation of muscle forces.

2. Methods

The method can be divided into the following steps:

(1) Create a model of lower extremity, choose segments and joints involved in the model and take account of the simplifications of their properties and functions.

(2) In an experimental measurement, determine the kinematics variables of the segments of lower extremity for one revolution of crank. Measure simultaneously the force of foot on the crank depending on the crank angle.

(3) Using the method of releasing and Newton–Euler's equations compute the reactions and joint torques in individual joints of mechanism.

(4) Using static optimization as well as physiological and morphological parameters of chosen muscles compute the forces in muscles.

2.1. 3D model

Our 3D model of the lower limb consists of seven segments (left foot, left shank, left thigh, pelvis, right thigh, right shank, and right foot), six joints (left and right hips, knee and ankle) and 36 muscles of each leg. The assumption was that the segments are rigid bodies, characterized by mass, length and the moments of inertia; joints are ideally spherical joints (without friction and clearance) and the only structures that are involved in the transfer of forces in the joint are muscles, tendons and the bone contact area as ZATSIORSKY [10] suggested.

2.2. Experimental measurement of kinematics variables

The Qualisys Motion Capture System was used for the record of the movement. The experiment was carried out in the Laboratory of Biomechanics of Extreme Loading in the Faculty of Physical Education and Sport, Charles University in Prague. The trajectories obtained were later modified by Qualisys Track Manager (QTM) software. Desired variables were subsequently obtained using the Visual3D Movement Analysis Software (C-Motion, Inc.). One particular cycling pattern at constant frequency and performance was chosen. It was not necessary to make more measurement with different load or frequency since the influence of these parameters has already been explored [11]–[14]. The measurement of the crank loading was necessary for the computation of dynamic parameters.

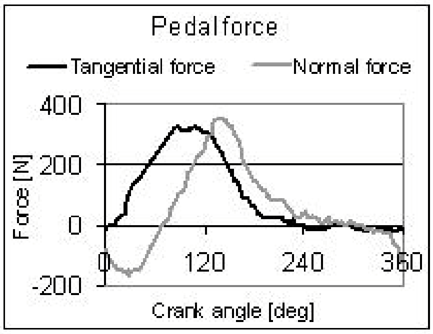

Fig. 1. Tangential and normal forces affecting pedal during one revolution of crank. The positive direction of the tangential force is in the direction of rotation of the crank; the positive direction of normal force is pointing from the centre of crank. The zero angle is in the top dead centre of the crank cycle

Reactions and moments are set up as parallel with the global coordinate system. The global x-axis leads form back to front, the global y-axis leads from right-hand side to left-hand side, and the global z-axis is vertical (the dextrorotary coordinate system is used). The beginnings of local coordinate systems are located in the centre of gravity of segments; the local z-axis coincides with the longitudinal axis of segment and leads from the distal end to the proximal end. The

x-axis is perpendicular to the z-axis, lies in the sagittal plane and makes the dextrorotary coordinate system with the y-axis.

The measured pedal force can be seen in figure 1. The maximal tangential and normal forces were reached near the first quarter of crank cycle. The starting point of the cycle is in the top dead centre. The negative sign in the tangential force means that the force is actually directed against the movement. The cyclist's performance can be improved by eliminating this stage. Moreover, such muscles as m. soleus, m. gastrocnemius caput mediale et laterale and other muscles that are involved in knee flexion and the stabilization of foot arch should reach their maxima in this phase.

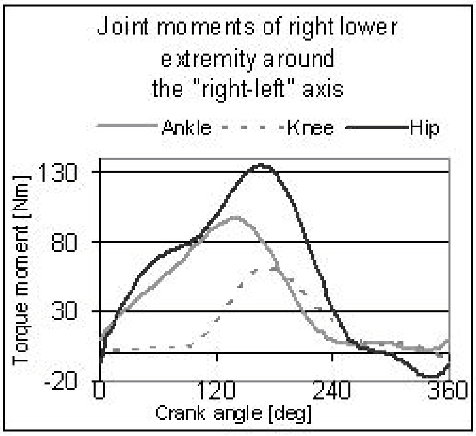

Fig. 2. Joint moments of right lower extremity around the local y-axis during one revolution of crank. The zero angle is in the top dead centre of the crank cycle

Using the pedal force and the kinematics variables, the moment of each joint was computed (figure 2). Since the main motion during pedalling is in the sagittal plane (the x–z plane), the torque moments in individual joint around the y-axis are most contributive to the muscle forces. Hence we were concentrated on the sagittal plane motion only.

2.3. The computation of the contact forces and torque moments

Weight, length, location of centre of gravity and matrixes of inertia of individual segments were computed using the equations and coefficients proposed by ZATSIORSKI [10] and De LEVA [15] with the following initial values: body mass = 65 kg, body high = 175 cm, gender: man. Contact forces and torque moments in individual joints were solved using Newton–Euler's equations

$$\vec{F}_v = \sum_i \vec{F}_i = m\vec{a}_S \,, \tag{1}$$

$$\vec{M}_{Sv} = \sum_i \vec{M}_i = \underline{I}_S \vec{\alpha} + \vec{\omega} \times \underline{I}_S \vec{\omega} \,, \tag{2}$$

where: m is the mass of the segment (kg), $\vec{a}_S$ is the acceleration of the centre of mass of the segment (m.s^{-2}), $\vec{F}_i$ are the forces affecting the segment (N), $\vec{F}_v$ is the resultant force affecting the segment (N), $\underline{I}_S$ represents the matrix of inertia of the segment (related to the centre of mass) (kg.cm^{-2}), $\vec{\alpha}$ is the angular acceleration of the segment (rad.s^{-2}), $\vec{\omega}$ is the angular velocity of the segment (rad.s^{-2}), $\vec{M}_i$ are the moments affecting the segments (Nm) and $\vec{M}_{Sv}$ is the resultant moment (Nm).

It is possible to write six scalar equations (1) and (2) for each segment that can be transformed into the matrix form. While we know the reaction and moments in the distal joint, we can easily compute the reaction and moments in the proximal joint of the segment. The above defined problem was solved in the program Matlab (MathWorks©, Inc.). These forces and moments in joints are the input parameters in the next step of computation.

2.4. Musculotendon dynamics

For the purpose of this work, the Hill-type model of the musculotendon complex was used [16]. The main parts of this muscle model are the passive component, which consists of an elastic element and passive muscle viscosity, and the active contractile component. The model for the active contractile component is based on the generally accepted notion that the active muscle force is the product of three factors: (1) the length–tension relation $f_L(L^M)$, (2) the velocity–tension relations $f_v(\dot{L}^M)$ and (3) the activation level $a(t)$. The muscle parameters necessary for creating scaled curves describing the attributes of generic muscle are: the maximum isometric active muscle force F_0^M, the optimum muscle length L_0^M, the pennation angle α_0 (when $L^M = L_0^M$) and the tendon slack length L_s^T. The model was in detail described by VILIMEK [17]. Then, the muscle force can be then calculated as follows:

$$F^M = [F_0^M . f_v(\tilde{v}^M) . f_L(\tilde{L}^M) . a(t) + F^{PE}] \cdot \cos\alpha \,, \tag{3}$$

the input arguments were found in [1], [18] and [19]. The following notation is used: F_0^M – the maximum isometric muscle force, α_0 – the pennation angle, L_s^T – the tendon slack length and L_0^M – the optimum muscle length. The only unknown variable in the equation is the activation level of the muscle $a(t)$, being solved using static optimization.

2.5. The distribution problem and static optimization method

If a system contains more unknowns than equations that describe a given mechanical system, it is said to be underdetermined, and in general, there is an infinite number of possible solutions. The distribution problem is used to solve internal forces acting on the musculoskeletal system using the known resultant joint forces and moments.

The kinematics variables and moments in joints are known from the experiment. What is more, the muscle contractions are supposed to be quasi-static, which means that the muscle forces depend on their current excitation only. So, the optimization can be solved independently in every moment. The problem is then solved as a minimization of cost function. Usually, the optimization problem is defined by three quantities: (1) the cost function, (2) the designed variables, (3) the constraint function. There are several approaches how to define the cost function and the optimization criteria. A wide review can be found in TSIRAKOS [8]. In this work, the function to be minimized (the cost function ϕ) was chosen as in VILIMEK [17]:

$$\phi = \sum_{i=1}^{N} (F_i^m)^2 , \tag{4}$$

or

$$\phi = \sum_{i=1}^{N} (a_i)^2 , \tag{5}$$

depending on the method used.

The notation is as follows: ϕ is the cost function, F_i^m is the force of the i-th muscle (N), N is the total number of the muscles examined; a_i is the activation level of the i-th muscle.

The designed variables (the muscle forces or the activation level) are systematically changed throughout the computation until the cost function is optimized and all constraint functions (equations (6), (7) and (8)) are satisfied.

$$F_i^m \geq 0 \quad \text{for} \quad i = 1, ..., N, \tag{6}$$

$$0 < a(t) \leq 1 \quad \text{for} \quad i = 1, ..., N, \tag{7}$$

$$M = \sum_{i=1}^{N} (r_i \times F_i^m) . \tag{8}$$

The drawback of the static optimization method with the inverse dynamics is its sensitivity to the quality of input (kinematics) variables [20].

2.6. Calculation without the consideration of physiological muscle parameters

In this case, the minimization of the sum of the muscle forces to the power of two (equation (4)) is found. It is necessary to satisfy the equality (equation (8)) and inequality (equation (6)) constraints. It is important to highlight that neither the activation level of the muscles, nor the physiological parameters of muscles are taken into consideration. The unknown variables are directly the forces in the muscles F^m, where r_i and M are known.

2.7. Calculation with the use of muscle physiological parameters

We took into consideration the activation level of muscles and their physiological parameters. The aim was to minimize the cost function (equation (5)), i.e. the sum of the activation levels to the second. The equality (equation (8)) and inequality (equations (6) and (7)) constraints must be satisfied. The F^M is computed from equation (3). The unknown variables are the activation levels of muscles $a(t)$. After computing these activations, the muscle forces can be computed as well (equation (3)).

3. Results and discussion

When we created our lower extremity model with the consideration of muscles, we made several simplifications relating to individual segments and joints. Firstly, segments are supposed to be rigid bodies, characterized by weight, length and the moments of inertia. Secondly, joints are said to be ideal spherical kinematics joints without friction and clearance. Finally, the only structures involved in the transfer of forces in the joint are muscles, tendons and the bone contact area.

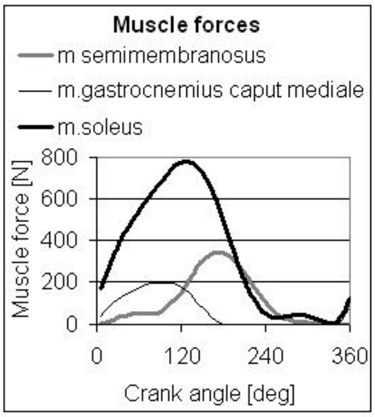

Fig. 3. Chosen muscle forces of lower extremity computed using the method without consideration of physiological muscle parameters. The zero angle is in the top dead centre of the crank cycle

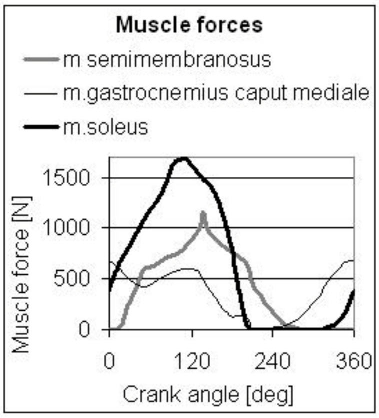

Fig. 4. Chosen muscle forces of lower extremity obtained using the method with consideration of the muscle activation and physiological parameters of muscles. The zero angle is in the top dead centre of the crank cycle

Muscle forces computed using the method without the consideration of physiological muscle parameters are shown in figure 3, and muscle forces obtained using the method with the consideration of the muscle activation and with physiological parameters of muscles are shown in figure 4.

In figures 5, 6 and 7, muscle forces in chosen muscles computed by means of both methods are presented. The forces obtained must be critically evaluated in relation to both the real motion and the maximum possible force in the muscles.

The major difference in the resultant force is in the two-joint muscles, e.g. m. gastrocnemius caput mediale (figure 6). Since this type of muscle acts in two different joints, i.e. it is involved in two different stages of crank cycle, the behaviour of the force during the cycle must have two maxima. This can be seen only based on the results of the method with the consideration of the physiological parameters of muscles.

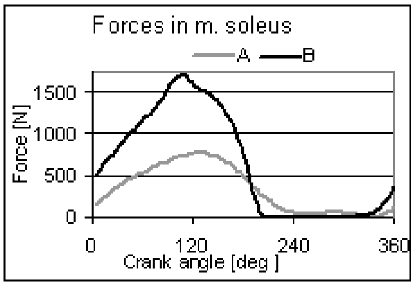

Fig. 5. Force in the m. soleus. The function of m. soleus is the plantar flexion of foot. A – neither the activation level of muscles nor the physiological muscle parameters were used. B – forces were computed with consideration of the activation level of muscles and physiological muscle parameters. The zero angle is in the top dead centre of the crank cycle

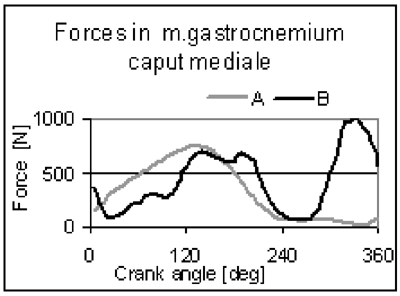

Fig. 6. The waveform of the force in the m. gastrocnemius caput mediale. The function of this muscle is the plantar flexion of foot and the auxiliary flexion of knee. The zero angle is in the top dead centre of the crank cycle. A – neither the activation level of muscles nor the physiological muscle parameters were used. B – forces were computed with consideration of the activation level of muscles and physiological muscle parameters

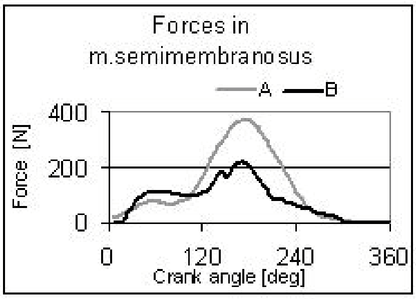

Fig. 7. The waveform of the force in the m. semimembranosus. Its function is the plantar flexion of knee and the auxiliary extension of the hip joint. The zero angle is in the top dead centre of the crank cycle. A – neither the activation level of muscles nor the physiological muscle parameters were used. B – forces were computed with consideration of the activation level of muscles and physiological muscle parameters

The behaviour of the force in one-joint muscles is more or less the same. The only difference in the behaviour of the muscle forces is in the magnitude of the force, even though the input variables are for all the computations the same. It is clear that the difference is caused by the choice of the cost function and by the use of the physiological muscle parameters.

4. Conclusion

From the dependence of muscle forces on the crank angle we can conclude that for the realistic determination of muscle forces during pedalling the physiological parameters of muscles must be incorporated in the computations. Only the equations with considering muscle physiology reflect reliably both maxima of the forces of two-joint muscles. Even though the results are sensitive to the incoming physiological parameters whose values differ a lot in the literature, it is still a contributing method.

Acknowledgement

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