

ASSESSMENT OF MUSCLE FORCES AND JOINT REACTIONS IN LOWER LIMBS DURING THE TAKE-OFF FROM THE SPRINGBOARD

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Abstract: Computer simulation methods, based on the biomechanical models of human body and its motion apparatus, are commonly used for the assessment of muscle forces, joint reactions, and some external loads on the human body during its various activities. In this paper a planar musculoskeletal model of human body is presented, followed by its application to the inverse simulation study of a gymnast movement during the take-off from the springboard when performing the handspring somersault vault on the table. Using the kinematic data of the movement, captured from optoelectronic photogrammetry, both the internal loads (muscle forces and joint reactions) in the gymnast's lower limbs and the external reactions from the springboard were evaluated. The calculated vertical reactions from the springboard were then compared to the values assessed using the captured board displacements and its measured elastic behaviors.

Key words: Musculoskeletal Human Models, Inverse Dynamics Simulation, Muscle Forces, Joint Reactions

1. INTRODUCTION

In studying human movements, the non-invasive experiments are usually limited to photogrammetry from which the positions and orientations of the body segments are captured, electromyography (EMG) used to record the sequence and timing of muscle activity, and measurements of the ground reaction forces. Direct recording of muscle forces and joint reactions in vivo are currently infeasible. In this situation, inverse dynamics simulation, based on human body modeling and the non-invasive measurements, is still the prevailing method for the assessment of the internal loads during various human activities.

The inverse dynamics methodology, aimed at the determination of muscle forces and joint reactions in the motion apparatus, can be divided into four main stages:

1. design of the physical (musculoskeletal) model of the human body and its motion apparatus;
2. formulation of the mathematical model;
3. capturing the movement kinematic data and (possibly) other experimental data;
4. calculations using appropriate numerical codes.

2. PHYSICAL MODEL

The gymnast body is modeled as a planar kinematic structure composed of $N = 16$ rigid segments (head, 3 trunk parts, arms, forearms, hands, thighs, legs, feet) interconnected by $k = 15$ ideal hinge joints. The motion of the segments is assumed to be represented in the sagittal plane. In the deterministic model of actuation, the interaction between the segments is modeled by means of k resultant muscle joint torques $\tau = [\tau_1 \dots \tau_{15}]^T$ and $l = 2k$ Lagrange multipliers $\lambda' = [\lambda'_1 \dots \lambda'_{30}]^T$ that represent the X and Y components of the joint reactions (Fig. 1c). In the non-deterministic model of actuation (Blajer et al., 2007, 2010), the

three control torques in the joints of each lower limb are replaced with the action of $m = 9$ lower-limb muscles and groups of muscles that actuate the three degrees of freedom (Fig. 1b), $F = [F_1 \dots F_9]^T$. In this model, due to the control overactuation in the lower limbs, the problem of distribution of the respective muscle torques τ' into the muscle forces F has infinite solutions, and is usually solved using optimization techniques (Winter, 2005; Winters and Woo, 1990; Yamaguchi, 2001). Applying the obtained muscle forces, the joint reactions in the lower limb will involve the tensile muscle forces, and as such λ obtained this way will differ from λ' obtained using the deterministic model.

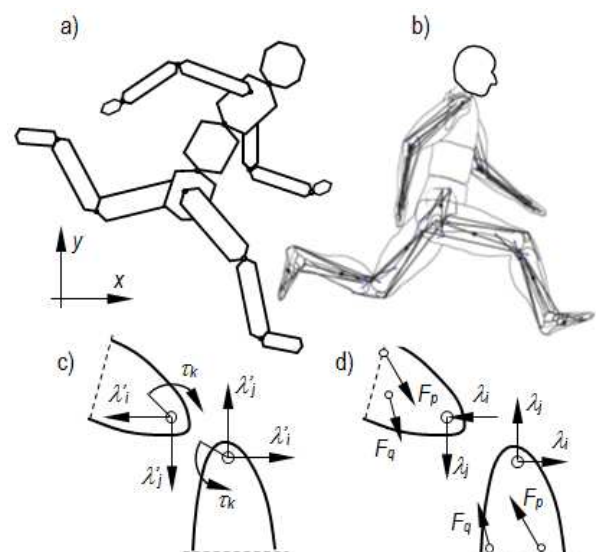


Fig. 1. The deterministic (a) and non-deterministic (b) models of the gymnast's body, and the interaction between the segments in the deterministic (c) and nondeterministic (d) models

The external loads on the gymnast's model are the gravitational forces $f_g = [f_{g1} \dots f_{g16}]^T$ of the respective segments, and the ground reaction forces reduced to a chosen point on the foot segments, $R = [R_x \ R_y \ M_z]^T$. A symmetric distribution of the ground reaction between the two legs during the phase of contact with the springboard was assumed.

The inertial and anthropometric data of the subject body, i.e. the segment masses and mass moments of inertia, the segment lengths and mass center locations, the cross-sectional areas of the modelled muscles, the effective origin and insertion points of the muscles and their paths relative the skeleton, etc., were either directly measured or estimated using the guidelines reported in the literature (Winters and Woo, 1990, Zatsiorsky, 2002, Tejszerska et al., 2011).

3. MATHEMATICAL MODEL

The dynamic equations of the gymnast model are formulated in $n = 3N = 48$ coordinates $p = [x_{C1} \ y_{C1} \ \varphi_1 \dots \ x_{CN} \ y_{CN} \ \varphi_N]^T$ that specify the location of the segment mass centers and their angular orientations with respect to an inertial (absolute) reference frame. The generic matrix form of the equation is:

$$M \ddot{p} = f_g + B_u u + C_\lambda \lambda^* + C_R R \quad (1)$$

As said, in the deterministic model, the internal loads are modeled by means of the resultant muscle torques in the joints, distributed here into the lower-limb joint torques τ' and the upper body joint torques τ'' , and the joint reactions λ' to which the contribution of the tensile muscle forces is not involved:

$$B_u u = B_\tau \tau = [B_{\tau'} \ | \ B_{\tau''}] [\tau' \ | \ \tau'']^T \quad \lambda^* = \lambda' \quad (2)$$

Then, in the nondeterministic model, the lower-limb joint torques τ' are replaced with the respective muscle forces F , which yields also more realistic joint reactions in the lower limbs (contribution of the tensile muscle forces is involved),

$$B_u u = [B_F \ | \ B_{\tau''}] [F \ | \ \tau'']^T \quad \lambda^* = \lambda \quad (3)$$

In the deterministic model, using the kinematic characteristics of the analyzed movement, the unknown internal loads and the external reactions can explicitly be determined from:

$$[\tau' \ | \ \tau'' \ | \ \lambda' \ | \ R]^T = [B_\tau \ | \ C_\lambda \ | \ C_R]^{-1} (M \ddot{p} - f_g) \quad (4)$$

The indeterminate inverse dynamics problem, i.e. the distribution of τ' from the determinate inverse dynamics formulation (4) into the respective muscle forces F , and then the determination of the joint reactions λ that include the influence of the tensile muscle forces in the lower limbs, can then be solved using the projected dynamic equations. These two aims are achieved by introducing $r = 18$ independent coordinates $q = [x_G \ y_G \ \varphi_1 \dots \ \varphi_{16}]^T$, where x_G and y_G are the absolute coordinates of a point on the top of the head segment, and the angular coordinates are as used in p . Then, using a relationship $p = g(q, z, t)$, where z are the open-constraint coordinates in the directions of λ^* , two matrices can be extracted from:

$$\dot{p} = D \dot{q} + E \dot{z} + \gamma \quad (5)$$

where the $n \times r$ (48×18) matrix D is an orthogonal complement

to the constraint matrix C_λ , $D^T C_\lambda^T = 0$, and the $n \times r$ matrix E its pseudo-inverse, $E^T C_\lambda^T = I$ (identity matrix). The equations (1) projected in q directions can then be represented as:

$$D^T (M \ddot{p} - f_g - C_R R) = D^T \frac{[B_{\tau'} \ | \ B_{\tau''}] [\tau' \ | \ \tau'']^T}{[B_F \ | \ B_{\tau''}] [F \ | \ \tau'']^T} \quad (6)$$

with the joint reactions excluded from the evidence.

Using the notation: $\bar{B}_\tau = D^T [B_{\tau'} \ | \ B_{\tau''}]$, $\bar{B}_{F\tau} = D^T [B_F \ | \ B_{\tau''}]$, and $\bar{M} = D^T M D$, after another projection of equations (6) into the controlled directions, one arrives at:

$$\bar{B}_\tau \bar{M}^{-1} \bar{B}_\tau \begin{bmatrix} \tau' \\ \tau'' \end{bmatrix} = \bar{B}_\tau \bar{M}^{-1} \bar{B}_{F\tau} \begin{bmatrix} F \\ \tau'' \end{bmatrix} \quad (7)$$

from which one can state the following relationship between the resultant muscle torques τ' and the muscle forces F (or stresses σ) in the lower extremities, i.e.

$$S_F F = P \quad \text{or} \quad S_F A_0 \sigma = P \quad (8)$$

where $S_F = (D^T B_\tau)^T \bar{M}^{-1} D^T B_{F\tau}$ is a square matrix of the dimension of F , $P = (D^T B_\tau)^T \bar{M}^{-1} D^T B_\tau \tau'$, and A_0 is the diagonal matrix of the muscle cross-sectional areas. The muscular load sharing problem in the lower-limbs can then be stated as the following optimization procedure

$$\begin{cases} \text{minimize} & J(\sigma) \\ \text{subject to} & S_F A_0 \sigma = P \\ \text{and} & \sigma_{\min} \leq \sigma \leq \sigma_{\max} \end{cases} \quad (9)$$

where J is an appropriate objective function, and σ_{\min} and σ_{\max} are the physiologically allowable minimal and maximal muscle stresses. In this way the lower-limb muscle torques τ' , obtained from the deterministic solution, are distributed into the respective muscle stresses/forces.

Using the evaluated muscle forces in the lower limbs, the joint reactions λ can be determined from the projection of the dynamic equations (1) into the constrained directions specified by E , i.e.

$$E^T M \ddot{p} = E^T \left(f_g + [B_F \ | \ B_{\tau''}] \begin{bmatrix} F \\ \tau'' \end{bmatrix} + C_\lambda \lambda + C_R R \right) \quad (10)$$

which, after using $E^T C_\lambda^T = I$, leads to

$$\lambda = E^T \left(M \ddot{p} - f_g - [B_F \ | \ B_{\tau''}] \begin{bmatrix} F \\ \tau'' \end{bmatrix} - C_R R \right) \quad (11)$$

4. SIMULATION RESULTS

The analysis was limited to the gymnast (master competitor of age 27, mass 70 kg, height 169 cm) movement during his landing and take-off from the springboard, with some short flying periods before and after the on-board phase, who performed a handspring vault with a front somersault in pike position (Fig. 2). The on-board phase was chosen for the expected high impact/impulsive forces and possible verification of the calculated external reactions from the springboard, which can be compared to the values assessed using the captured board displacements

and its measured elastic behaviors.

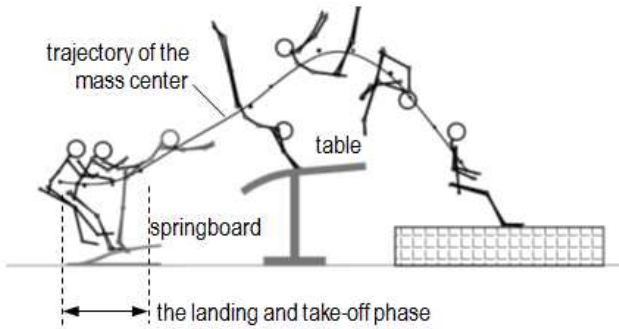


Fig. 2. Distinct phases of the performed handspring vault with a front somersault in pike position

The actual jump performance was recorded using a set of synchronized digital cameras (100 Hz), together with a separate registration of the springboard displacements. The raw kinematic data were then smoothed (Winter, 2005; Dziewiecki et al., 2011) to obtain the base point trajectories, from which the kinematic characteristics $p_d(t)$, $\dot{p}_d(t)$ and $\ddot{p}_d(t)$, used in the inverse simulation study, were calculated.

In Fig. 4 are reported the simulation results for the muscle forces of four selected group of muscles: *r.fem.* (*rectus femoris*), *vast.* (*vastus*), *gastr.* (*gastrocnemius*), and *sol.* (*soleus*), presented in Fig. 3.

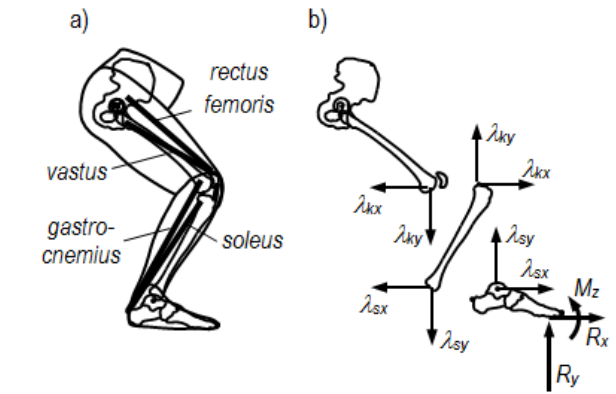
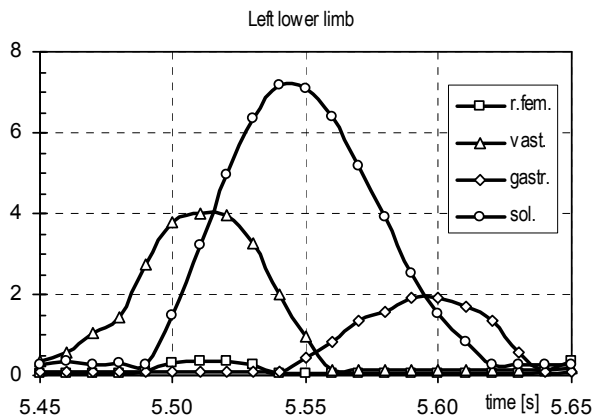


Fig. 3. The selected groups of muscles (a) and the joint reactions in lower limbs together with the external reactions (b)

As shown, the results obtained for the right and left limbs are very similar. During the on-board phase, which lasts from 4.48 to 5.58 s, the largest contribution to the movement performance comes from *sol.*, the contribution of *r.fem.* appears meaningless, *vast.* is active mainly during the landing, and *gastr.* during the take-off. Then, in Fig. 5 there are reported the calculations of the knee and ankle reactions during the movement. The estimated total reaction values, respectively in the knee (k) and ankle (s) joints are, are approximately 7 and 10 times higher than the gymnast weight.

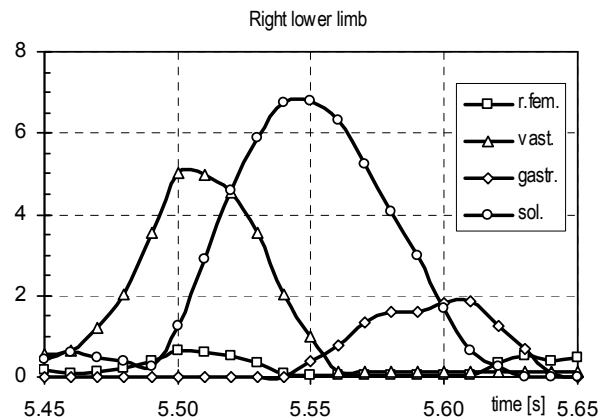


Fig. 4. The forces in selected muscles of the left and right lower limbs during the take-off from the springboard, related to the body weight of the gymnast

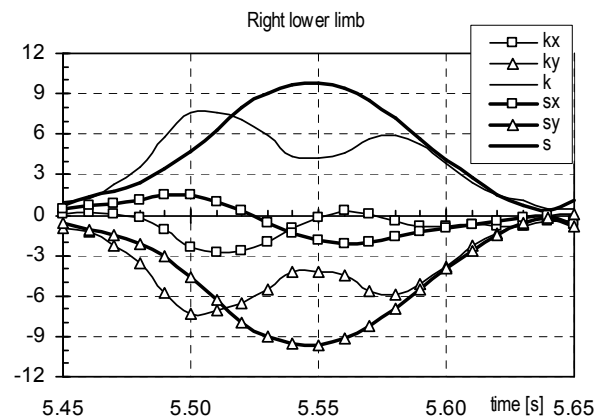
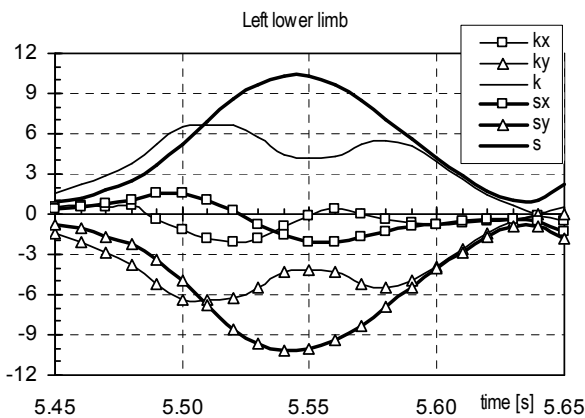


Fig. 5. The reactions in the knee (k) and ankle (s) joints of the left and right lower limbs during the take-off from the springboard, related to the body weight of the gymnast

5. LIMITED VALIDATION OF THE SIMULATION RESULTS

In addition to the calculations, the vertical reactions R_y from the springboard during the take-off were estimated using the captured board displacements and its experimentally measured elastic characteristics (Mazur et al., 2011b). The maximum values of the vertical reaction R_y , calculated from the inverse dynamics analysis and estimated from the recorded board displacements, were in good agreement (Fig. 6). This allows one to have limited confidence to the quality of the developed biomechanical model, correctness of its geometric and inertial parameters, and, finally, accuracy/adequacy of the kinematic characteristics used as an input to the human dynamics model in the inverse dynamics simulation (Erdemir et al., 2007).

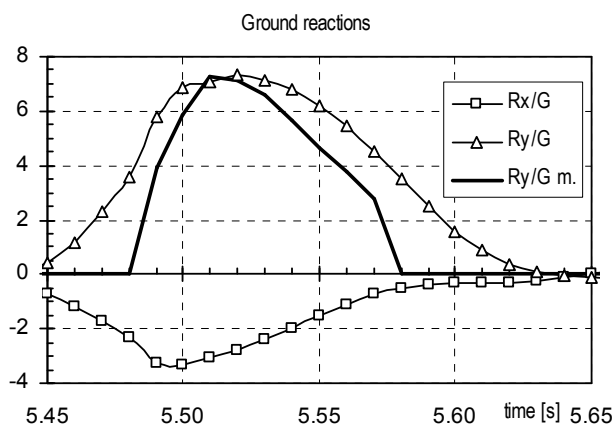


Fig. 6. The horizontal and vertical reactions from the springboard obtained from the inverse dynamics simulation, and the vertical reaction assessed from the recorded board displacements, related to the body weight (G) of the gymnast

6. CONCLUSIONS

The human motion apparatus is extremely complex and, as such, very difficult to model. For these reasons the models used in the inverse dynamics analyses always involve simplifications, according to the aims and expected exactitude of the analysis.

Studies on muscle force prediction usually compare the assessed muscle force loading or activation patterns against the EMG data as an estimate of validity. In this paper we compared the springboard vertical reaction values obtained from the inverse simulation against the values estimated from the measured board displacements. A good agreement of the maximum vertical values was reported.

In the literature, see e.g. Erdemir et al. (2007) for a review, advanced analyses exist which incorporate the quantification of muscle force sensitivity on diverse modeling parameters. Some critical model parameters are associated with the assumptions related to the musculotendon paths and the effective attachment points of the tendons (Winters and Woo, 1990; Blajer et al., 2010). The physiological cross-sectional area of muscles are the other parameters that significantly affect the magnitude of muscle force

estimates (Mazur et al., 2011a). Of importance is also the way the raw kinematic data are processed (smoothed/filtered) before they are used in the inverse dynamics simulation (Dziewiecki et al., 2011). Finally, the muscle force estimates are influenced by muscle decomposition and recruitment criteria used in the force sharing optimization process. Nonetheless, though the inverse dynamics simulations are possibly burdened with possible large inaccuracy, they still remain the only prevailing non-invasive method for the assessment of the internal loads during human movements.

The reported evaluations show that in gymnastics, during the dynamic movements like landing and take-off, the internal loads in the lower limbs may be much (a dozen or so) higher compared to the gymnast weight. The situation is reflected in frequent injuries of limb structures of locomotion apparatus, and different diseases after longer sport activity. Knowledge of muscle forces and joint reactions during the sport activities, even if approximate, can be of great importance for the risk assessment.

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