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IDENTIFICATION OF INTERNAL LOADS AT THE SELECTED JOINTS DURING PERFORMANCE OF A BACKWARD SOMERSAULT

A backward tucked somersault from the standing position is analyzed in this study. The used computations were based on a three-dimensional model of the human body defined in natural coordinates. The net torques and internal reactions occurring at the ankle, knee, hip, upper trunk-neck and neck-head joints were obtained by inverse dynamics. The achieved results confirmed the significant loads at the joints at the beginning of the landing phase and revealed the way the analyzed stunt was controlled. It was also shown that the natural coordinates provide a useful environment for modeling spatial biomechanical structures.

1. Introduction

A backward tucked somersault from the standing position is one of the most popular acrobatic jumps. Students are taught this stunt in basic gymnastic courses at faculties of physical education. Some football players or weight lifters make it spontaneously after scoring or achieving the record result. Quite frequent performances of this jump by athletes of different sport disciplines suggest that the backward somersault deserves biomechanical identification.

The identification, which is focused on obtaining the internal loads at the selected joints of the human body, has two main components. First, a three-dimensional model is used in the computations. Such biomechanical models have been existing for almost 50 years [1], and are still being developed [2]. Their use, however, with the exception of gait analysis, is rather seldom, particularly if the aim of the research is to estimate internal joint reactions, e.g. [3, 4].

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The other component creates a set of natural coordinates [5] defined to describe the kinematic structure of the biomechanical model. The first 3D human body model based on natural coordinates was employed for a dynamic analysis of gait, a football kick and some athletics events at the Barcelona Olympic Games [6, 7]. The other model [8], containing a muscle apparatus of the lower leg and using different approach for modeling feet and hands, has been used so far for gait and long jump analyses [9, 10]. The backward somersault gives an opportunity to validate this model in a movement of the human body that leads to the change of space positions of anatomical segments by a complete rotation of 360 degrees.

Thus, 3D identification of internal loads and validation of the second biomechanical model during a performance of the somersault are the key purposes of this work.

2. Biomechanical model

The biomechanical model shown in Fig. 1a consists of 16 anatomical segments. The model is described by natural coordinates used to construct its kinematic structure made of 33 rigid bodies branching from the pelvis in open-chain linkages. The rigid bodies that form the neck, arms, forearms, thighs, shanks, upper torso (numbers in circles from 19 to 25 in Fig. 1b) and the lower torso (numbers 6, 7, 8) are defined by the Cartesian coordinates of two points and one unit vector each. The hands, feet and the head are defined by the coordinates of three points and one unit vector. The complete set of rigid bodies is described by means of 25 points and 22 unit vectors (Fig. 1b), accounting for a total number of 141 natural coordinates.

The biomechanical model has revolute and universal joints only. The universal joints are located in the ankle, the radioulnar articulations, between the 12th thoracic and 1st lumbar vertebrae (designated by lower-upper torso joint), and between the 7th cervical and 1st thoracic vertebrae (referred to as the upper torso-neck joint). Since in this formulation most joints are defined naturally by sharing points and vectors between adjacent rigid bodies, the most common kinematic constraints that arise here are of the rigid body type. There are no joint constraints except universal joints, where the appropriate unit vectors ought to be kept perpendicular to each other. The model has 44 degrees-of-freedom that correspond to 38 rotations about the revolute and universal joints, and 6 degrees-of-freedom associated with the pelvis, which is treated as the base body. The degrees-of-freedom of the biomechanical model corresponding to rotations about the joints are numbered in Fig. 2a.

One way of calculating internal net torques at the joints of the biomechanical model is to assume that each degree-of-freedom is driven by

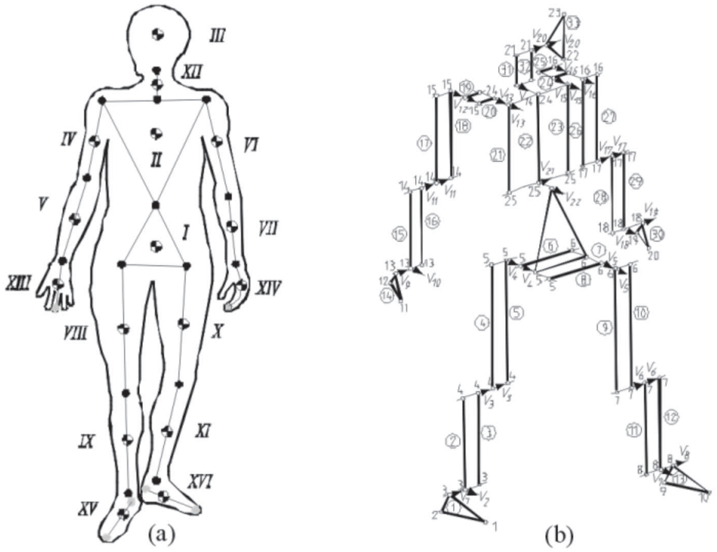

Fig. 1. Biomechanical model of the human body (left) and its kinematic structure (right)

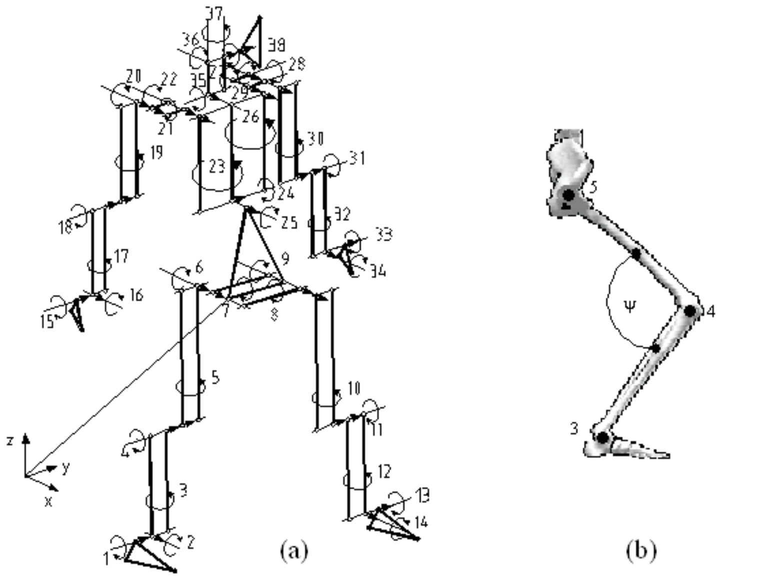

Fig. 2. Degrees-of-freedom of the model (left) and symbolic representation of appended driving constraints (right)

an adequate joint actuator [11], such as the one depicted in Fig. 2b. Joint actuators are introduced into the Jacobian matrix of the system by means of kinematic driver constraints of scalar product type

$$(\mathbf{r}_5 - \mathbf{r}_4)(\mathbf{r}_3 - \mathbf{r}_4) - l_{34}l_{45}\cos(\psi) = 0. \tag{1}$$

The angle Ψ must be known and its discrete values are calculated from the measured spatial positions of the basic points.

Finally, most of the main human segments are modelled with two coincident mechanical elements. Such modelling approach is typical for natural coordinates [5, 6], resulting from the requirement of representing the rotation of each anatomical segment with respect to itself (Fig. 3), and from the need of explicit formulation of constraints to calculate internal reactions in the joints.

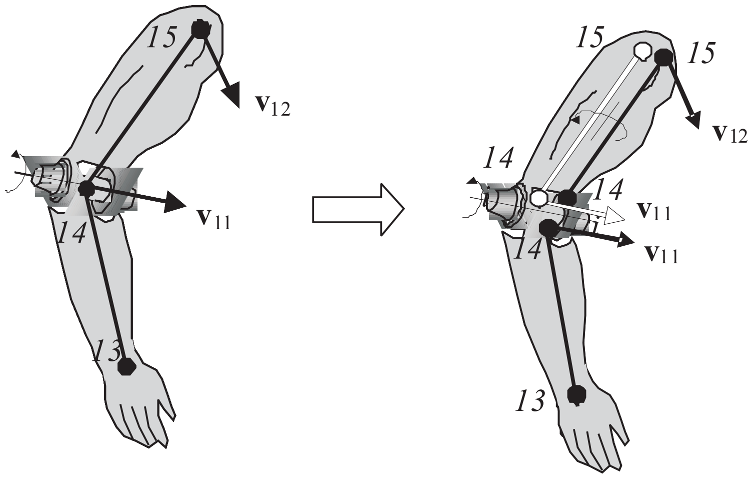

Fig. 3. Rotation of anatomical segment with respect to itself

Two different rigid bodies share points *14* and *15* in order to define the revolute joint that describes the medial/lateral rotation of the upper arm

$$\Phi_{19} = \mathbf{v}_{11}^T\mathbf{v}_{12} - \cos(\mathbf{v}_{11}, \mathbf{v}_{12}) = 0. \tag{2}$$

The angle between vectors $\mathbf{v}_{11}$ and $\mathbf{v}_{12}$ is derived just as angle Ψ from equation (1). The analytical expressions for these constraints as well as for all the appended driving constraints are presented in [12].

3. Inverse dynamics

Natural coordinates are dependent. The most common constraints originate from constant lengths of the rigid body links, constant angles between links and unit vectors and from unit modules of vectors. All the constraints, including the appended ones (Fig. 2b), can be expressed in the generic form

$$\mathbf{\Phi}(\mathbf{q}) = \mathbf{0}, \tag{3}$$

where vector $\mathbf{q}$ contains the coordinates of the basic points, the components of the unit vectors, and known angles.

Newton-Raphson's method is usually applied to solve the nonlinear system of constraints equations (3)

$$(\mathbf{\Phi}_q^T \mathbf{\Phi}_q)_i (\mathbf{q}_{i+1} - \mathbf{q}_i) = -(\mathbf{\Phi}_q^T)_i (\mathbf{\Phi})_i \tag{4}$$

where $\mathbf{\Phi}_q$ is the Jacobian matrix of the constraints, and i denotes iteration. A least square approach is required because the double rigid body's arrangement of most biomechanical segments generates additional constraints causing the total number of constraints to exceed the number of natural coordinates.

The next step is to calculate the velocities and accelerations of natural coordinates differentiating Eq. (3) twice with respect to the time

$$\mathbf{\Phi}_q \dot{\mathbf{q}} = \mathbf{0} \quad \Rightarrow \quad \mathbf{\Phi}_q \ddot{\mathbf{q}} = -\dot{\mathbf{\Phi}}_q \dot{\mathbf{q}} \ . \tag{5}$$

The dynamic equations of the motion for the biomechanical model described in natural coordinates can be written [5, 10, 11] as

$$\mathbf{M}\ddot{\mathbf{q}} = \mathbf{G} - \mathbf{\Phi}_q^T \boldsymbol{\lambda} \tag{6}$$

where $\mathbf{M}$ denotes the mass matrix of the system, $\ddot{\mathbf{q}}$ the vector of accelerations, $\mathbf{G}$ the vector of external loads containing the gravitational forces and reactions from the ground, and $\boldsymbol{\lambda}$ is the vector of Lagrange multipliers associated with the constraints forces and net driving torques occurring at the joints.

4. Data acquisition

A twenty-three-year-old adult male, with the height of 168 cm and a body mass of 68 kg, turned several back somersaults. One of the trials was chosen as the most representative and was subjected to the identification. In all the jumps the force-plate data and the body motion kinematic data were synchronized and recorded simultaneously. The ground reaction forces were measured by means of Kistler 9281B force platform at a sampling frequency at 1000 Hz, while the body motion was videotaped at 50 Hz by 4 synchronized cameras. The global reference frame was attached to the center of the force platform. The positions of the 23 anatomical points were digitized in APAS package environment to reconstruct the motion. Other steps of data acquisition process as well as handling the raw kinematic data were similar to those described elsewhere [13]. The placement of digitized points is illustrated by two pictures shown in Fig. 4, obtained during the take-off and airborne phase of the somersault.

Having been an elite acrobat several years ago, the jumper performed the standing backwards somersault extremely well with a well developed technique, as represented in Fig.4. The upper extremities and the lower ones were collateral during the stunt. The feet, having the same initial position in the anterior-posterior direction, were shifted only 0.8 cm apart from each other after the landing. Since external forces were measured with one force plate only, the correct technical performance of the jump allows us to presume the symmetrical distribution of the ground reactions for the feet. The application point of the ground reaction was established on the heel-metatarsal line being coincided with the cast of the center of mass of the body on the force plate in the anterior-posterior direction.

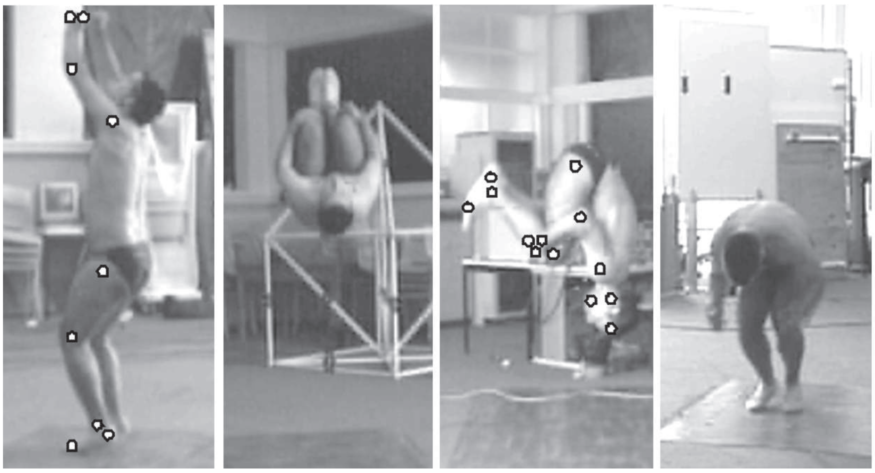

Fig. 4. Backward tucked somersault from the standing position, the placement of 23 digitized points is marked by circles

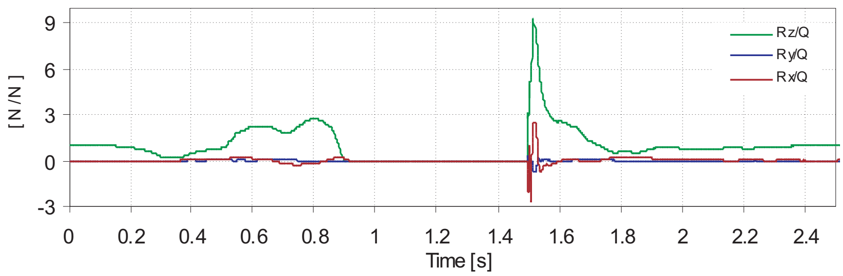

Fig. 5. Normalized (to body weight) components of ground reaction forces during the analyzed backward somersault

An impulsive character of the movement during the landing becomes apparent in internal load courses in the joint of the model. Fig. 5 shows, then, the normalized courses of component reactions from the side of the ground acting on the jumper's foot in this stage of the somersault. The vertical reaction assumes the values which are ninefold bigger than the body weight of the athlete. The level of the horizontal reaction is also high.

5. Results

The time courses of the component moments attributed to the first, fourth and seventh degrees-of-freedom of the model are presented in Fig. 6. They represent an action of flexion or extension in ankle, knee, and hip joints. All the characteristics possess a clearly emphasized phase of the take-off expanding to 0,9 s and the landing starting from 1,5 s. In both phases, the component moment in a hip joint assumes the highest values and its negative

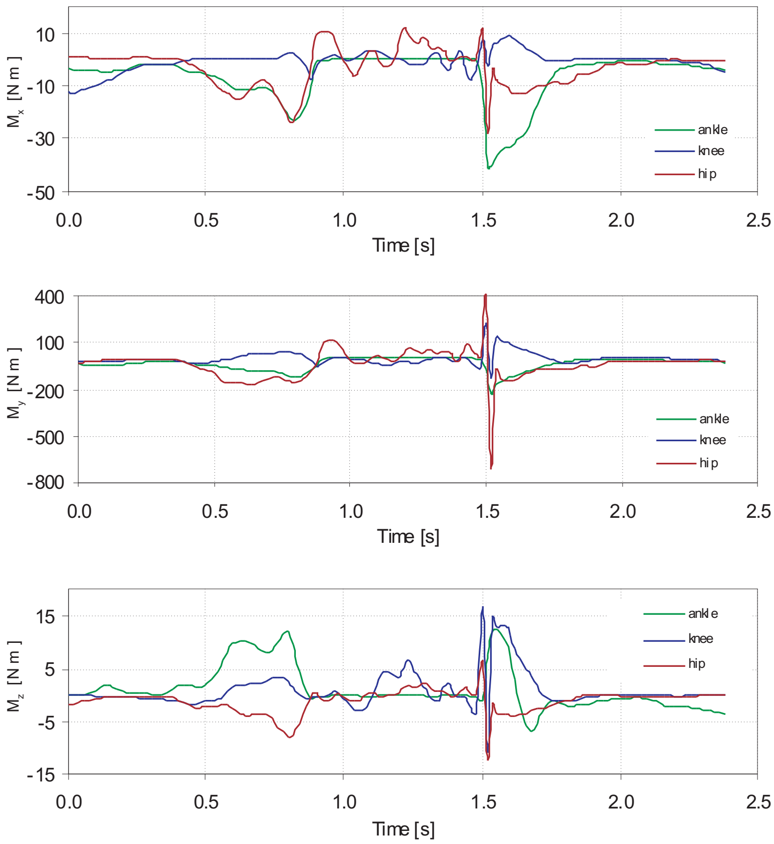

Fig. 6. Component moments associated with the 1st, 4th and 7th degrees-of freedom

value indicates some action from hip extensors. The sudden change of values of this moment from the level 400 Nm to – 700 Nm in the beginning of the landing reflects the impact of the ground reaction forces, and must be absorbed through the tendons and passive joint structures (ligaments), since the muscles are not able to react in such a short time to generate such a large moment. The moment courses in the sagittal plane during the flight phase have similar characteristics with their corresponding courses in a backward tucked somersault on the trampoline [14], whereas in the landing phase they are similar to the moments presented in the work [15].

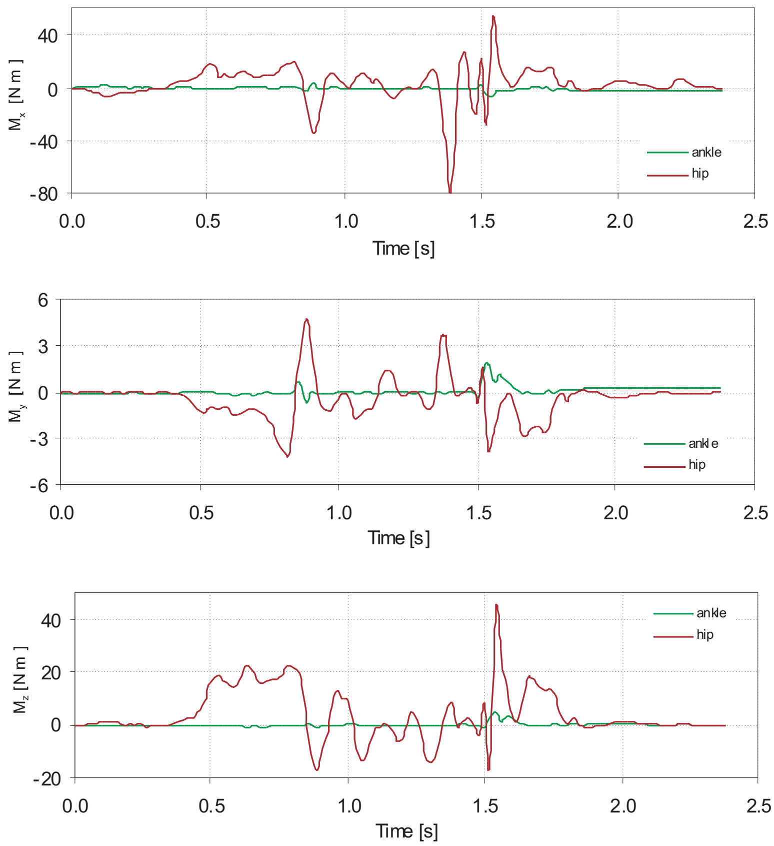

Fig. 7. Component moments associated with the 2nd and 6th degrees-of freedom of the biomechanical model

Figure 7 shows the characteristics of the component moments linked with the second and sixth degrees-of-freedom. They refer to the movements in the frontal plane. The load of the foot in the frontal plane is of small significance, but abduction and adduction torques in this plane obtain the moment values reaching –40 Nm at the end of the take-off and –80 Nm in the phase of the flight when the body starts disgrouping. The values of the component moments in the transversal plane in the final phase of the take-off and in the beginning of the landing also assume the same order of magnitude.

Figure 8 presents the component moments associated with the third and the fifth degrees-of-freedom of the biomechanical model. They correspond to the foot pronation and supination in the ankle joint as well as internal and external rotations in the hip joint. All the component moments have decidedly lower values than the ones described earlier.

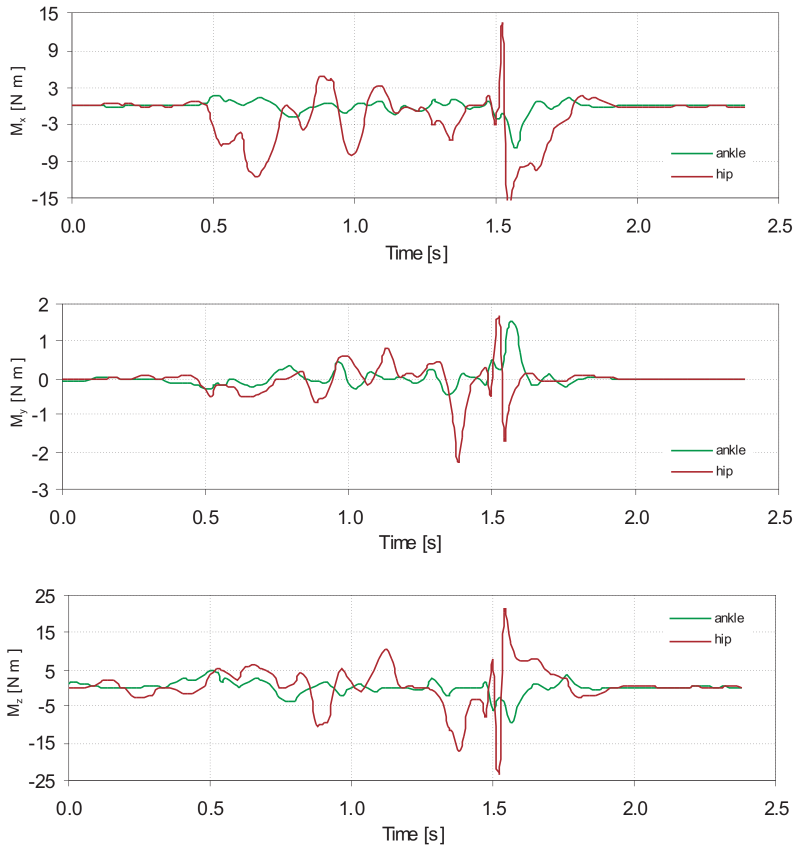

Fig. 8. Component moments associated with the 3rd and 6th degrees-of freedom of the biomechanical model

Figure 9 focuses on the characteristics of horizontal (R_x), medial-lateral (R_y), and vertical (R_z) reactions in the left leg joints taking place during the analyzed somersault. After Lagrange multipliers were calculated [5, 12], it was possible to estimate the reactions in the natural coordinates' environment. The figure shows the damping effect of the vertical reaction between the ankle joint and the hip joint. The highest value of this reaction in the landing moment reaches 3000 N in the ankle joint and about 1000 N in the hip joint. The shape of the vertical reaction makes it possible to identify explicitly the phases of countermovement (up to 0.5 s) and take-off (0.5 s – 0.9 s). Also, a variable course and the level of the horizontal reaction during the landing as well as the medial-lateral reaction in the flight phase from 0.9 s to 1.5 s is worth noticing. While comparing the values of reactions in the ankle joint with the reaction of the ground, it should be remembered that Fig. 5 presents the components of reactions acting at the same time for both legs.

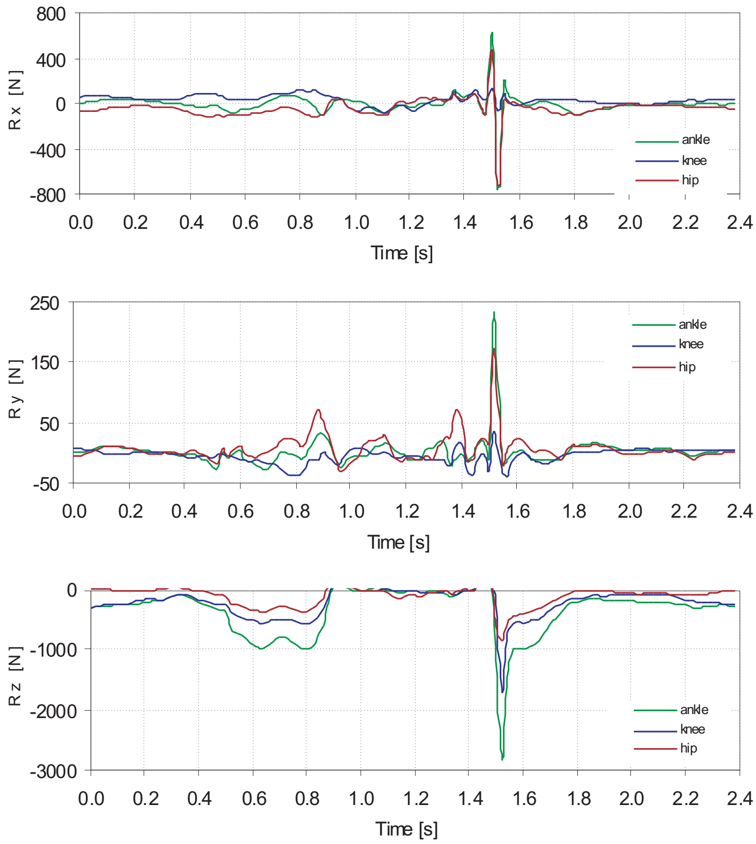

Fig. 9. Internal reactions at the joints of the right leg

The moment at the level of the connection between the head and the neck (Fig. 10) possesses two clear extremes in the sagittal plane. The first one occurs in the middle of the phase of increasing the load during the take-off when the head is bending back intensively, and the other one is caused by the head movement taking place in the middle of the flight phase in order to sustain eye contact with the ground. An unimportant influence of reaction forces on the analyzed characteristics during the landing results from the fact that the trunk and the head of the athlete are parallel to the ground at that time.

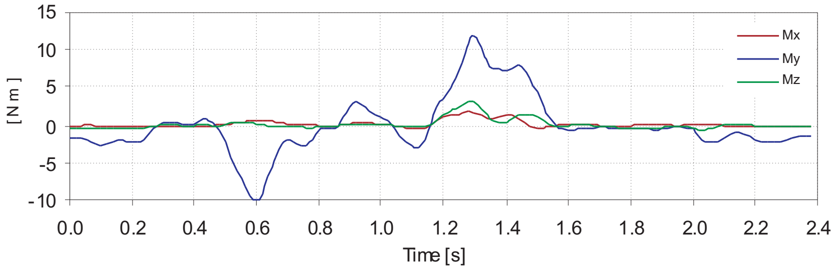

Fig. 10. Component moments at the head-neck joint

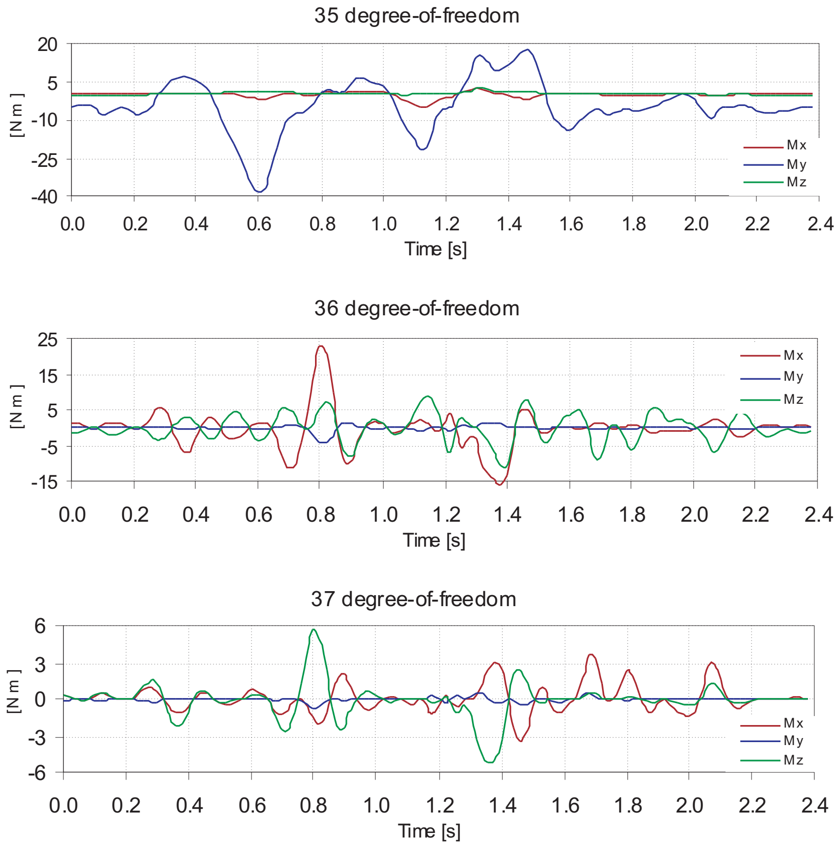

Fig. 11. Component moments in the basic anatomical planes on the level of the trunk-neck and neck-head joint

Figure 11 presents a rarely described in the literature on the subject three-dimensional state of loads in the place where the neck and the trunk are linked at the level of C7. The 35th degree-of-freedom reflects the neck movements with respect to the trunk in the sagittal plane, the 36th – side bending of the neck in the frontal plane, whereas the 37th one corresponds to the turn of the neck with respect to the trunk in the transversal plane. The highest values are assumed by the component moment connected with the 35th degree-of-freedom in the sagittal plane. The course of that moment is similar to the already discussed characteristics at the level of the neck-head connection and again, a significant domination over the others is clearly visible. The component moments connected with the 36th and the 37th degrees-of-freedom also possess the prevailing components of the torque in the sagittal and transversal planes, but they are accompanied by non-zero components in these planes.

A supplement to the contents of the last figure is a three-dimensional state of internal reactions on the level of the neck-head connection illustrated in Fig. 12.

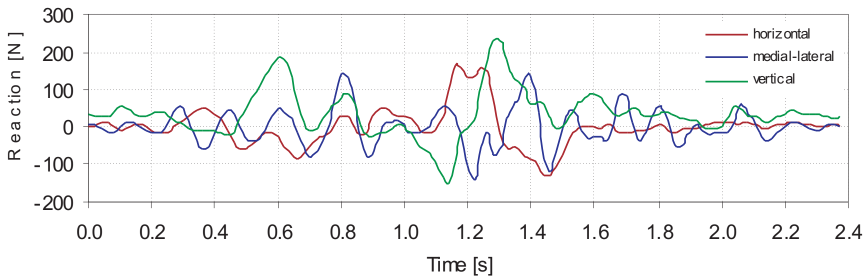

Fig. 12. Internal reaction components on the neck-head joint

When interpreting all the characteristics presented above one should be aware that inverse dynamic solutions in biomechanics are subject to systematic errors arising from the digitization process, improper filtering of the kinematic input data, inaccurate estimation of body segment dimensional and inertial parameters, as well as incorrect measurements of the external forces. The systematic errors may affect the accuracy of the calculations. In order to conduct a classic error analysis, an evaluation of the systematic errors is indispensable. The task proves to be complicated due to the above-mentioned factors causing the errors. On the other hand, the proposed methodology seems to be mathematically correct since it gives the same results in case of human gait analysis as the Newton-Euler approach [8].

The validation of the biomechanical model has been limited to the calculation of the component moments associated with some degrees-of-freedom of the biomechanical model, and internal reaction components at selected joints. Some of the computed characteristics have then been explained in terms of their relevance to the analyzed motor task. The validation itself of three-dimensional biomechanical model is, in general, a much more complex process, and the other aspects of this process are not addressed in the current study.

6. Conclusion

The work focuses on the loads' identification in leg and neck joints occurring during a backward tucked somersault from the standing position. The obtained time courses of torques and internal reactions in the joints have, apart from cognitive aspects, a clear interpretation, reflecting a distinctly impulsive character of movement in the landing phase and the peculiarity of the other phases of a somersault.

A three-dimensional model of the human body defined by natural coordinates turned out to be effective in full rotations of individual parts of the body. A space reconstruction of the jumper's movement based upon 23 measure points located in the joints is also worth noticing.

The comprehension of values and the load characteristics in the joints during the analyzed stunt can be of interest to both athletes and gymnastics trainers.

An analysis of a greater number of somersaults seems to be necessary in order to answer the question whether a proper body configuration in the beginning of the landing influences the loads' reduction in the joints. If so, the research will have a practical quality, since its results can help to reduce the risk of injuries.

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Identyfikacja obciążeń wybranych stawów podczas salta z miejsca w tył w pozycji kucznej

S t r e s z c z e n i e

W niniejszej pracy poddano identyfikacji biomechanicznej salto z miejsca w tył w pozycji kucznej. Obliczenia przeprowadzono wykorzystując trójwymiarowy model ciała człowieka zdefiniowany we współrzędnych naturalnych. W wyniku rozwiązania zagadnienia odwrotnego uzyskano przebiegi zbiorczych momentów sił mięśniowych oraz reakcji wewnętrznych w stawach kończyny dolnej i na poziomie połączeń tułowia z szyją oraz szyi z głową. Uzyskane wyniki wskazują na wysoki poziom obciążeń w stawach na początku fazy lądowania oraz ukazują sposób sterowania ciałem człowieka w trakcie badanego salta. Obliczenia potwierdziły też przydatność współrzędnych naturalnych w modelowaniu przestrzennych struktur biomechanicznych.