International Journal of Occupational Safety and Ergonomics

ISSN: 1080-3548 (Print) 2376-9130 (Online) Journal homepage: https://www.tandfonline.com/loi/tose20

Multibody model of the human upper extremity for fracture simulation

Marcin Milanowicz & Krzysztof Kędzior

To cite this article: Marcin Milanowicz & Krzysztof Kędzior (2016) Multibody model of the human upper extremity for fracture simulation, International Journal of Occupational Safety and Ergonomics, 22:3, 320-326, DOI: 10.1080/10803548.2015.1131070

To link to this article: https://doi.org/10.1080/10803548.2015.1131070

International Journal of Occupational Safety and Ergonomics (JOSE), 2016
Vol. 22, No. 3, 320–326, http://dx.doi.org/10.1080/10803548.2015.1131070

Multibody model of the human upper extremity for fracture simulation

Marcin Milanowicz* and Krzysztof Kędzior

Central Institute for Labour Protection – National Research Institute (CIOP-PIB), Poland

About 3.8 million people are injured in accidents at work in Europe every year. The resulting high costs are incurred by the victims themselves, their families, employers and society. We have used a numerical simulation to reconstruct accidents at work for several years. To reconstruct these accidents MADYMO R7.5 with a numerical human model (pedestrian model) is used. However, this model is dedicated to the analysis of car-to-pedestrian accidents and thus cannot be fully used for reconstructing accidents at work. Therefore, we started working on the development of a numerical model of the human body for the purpose of simulating accidents at work. Developing a new numerical model which gives an opportunity to simulate fractures of the upper extremity bones is a stage of that work.

Keywords: accident reconstruction; numerical simulation; multibody; MADYMO; human model

1. Introduction

In recent years, numerical methods have been developed rapidly. Numerical methods and their application are used to simulate various physical phenomena ranging from testing the strength of a structure to calculating the dynamics and kinematics of physical objects. The use of numerical methods to analyze the dynamics of physical bodies provided the opportunity to simulate accidents,[1] and thus the ability to reconstruct the course of events, their effects and causes, based on the laws of physics. Numerical methods are therefore an excellent tool for improving security, reducing the number of accidents and assisting in establishing liability in court. For these purposes, specific computer programs, e.g., MADYMO [2] or LSDyna,[3] are used. These programs include libraries with advanced numerical models of human bodies and crash test dummies. These models can be used to simulate human kinematics and dynamics and to assess the probability of injuries which a human body would suffer during an accident. Individual parts of the body are assessed with injury criteria. These criteria make it possible to associate the physical quantities recorded by 'virtual sensors' in the model with the injuries of a human body.[4]

Researchers at the Central Institute for Labour Protection – National Research Institute (CIOP-PIB) have worked on using numerical simulations to reconstruct accidents at work for several years now. This work focuses on selected events related to mechanical hazards such as falls, overturns or impact by objects.[5] To reconstruct these accidents, MADYMO R7.5 is used with a numerical model of the human body (pedestrian model) provided with the software. This model is dedicated to analyzing car-to-pedestrian accidents. However, it has some shortcomings, e.g., it is impossible to simulate fractures of the bones of the upper extremities. Therefore, CIOP-PIB started working on the development of numerical models of human body segments for the purpose of simulating accidents at work.[6] One of the stages consists of developing a numerical model of the upper extremity, which makes it possible to simulate bone fractures. The ability to simulate these injuries is very important in a numerical model dedicated to reconstructing accidents at work. Poland's Central Statistical Office (GUS) data show that upper extremity injuries occur in 40% of people injured in accidents at work in Poland.

This article describes a multibody (MB) numerical model of the upper extremity developed at CIOP-PIB. This model makes it possible to simulate fractures of the humerus, radius and ulna bones.

2. Methods

MADYMO makes it possible to create a model which combines two different numerical methods: MB and finite element (FE). The MB method is especially useful for simulating the motion of non-deformable bodies and their chains, whereas the FE method is commonly used for simulating kinematics and dynamics of deformable structures. The MB model consists of rigid bodies interconnected by kinematic joints. Properties such as masses, moments of inertia and center of gravity locations are defined for each body in the system. In order to simulate

*Corresponding author. Email: marmi@ciop.pl

real conditions of the modeled system, one should define external loads acting on the bodies (e.g., gravity), contact interactions between bodies, friction coefficients and/or stiffness characteristics. Differential equations describing motion are solved with numerical integration, resulting in displacements, velocities and accelerations of rigid bodies. These data, combined with masses and moments of inertia, are used to calculate forces and moments acting within the system. In MADYMO, contact between rigid bodies and FE models can be defined. Moreover, physical parameters such as damping and/or friction can be included in contact definitions. For objects created using both methods (MB and FE), contact forces prevent surfaces from going through each other. If a deformable structure is used, it is deformed in accordance with the properties of the material.[7]

FE simulation is more accurate and provides more complex results; however, it is much slower than MB. Typical FE simulations in crash test reconstructions take less than 250 ms; however, it takes a few or even a few dozen hours of CPU time to complete FE calculations. Reconstructions of accidents at work are much longer; they take up to 10 s. The MB method can produce results in more reasonable time. At the same time, MB is sufficiently accurate for accident at work modeling and simulations. Considering this, we decided to develop a model of the human upper extremity with the MB method.

3. Results

The model of the human upper extremity for simulating fracture was developed to reflect an upper extremity of the 50th percentile human (male) model. It consists of one segment modeling the arm, one segment modeling the forearm and one segment modeling the hand. In order to simulate fracture of the humerus, radius and ulna bones, the arm and forearm segments consist of two rigid bodies each. The arm segment consists of the upper part of the arm (Figure 1, 6) and the lower part of the arm (7). They are interconnected by a spherical joint with three degrees of freedom (2). The forearm segment also consists of the upper part of the forearm (8) and the lower part of forearm (9). These parts are also interconnected by a spherical joint with three degrees of freedom (4). Fractures are simulated with the use of these joints; they are described in Section 3.1.

Segments of the upper extremity (two bodies of arm, two bodies of forearm, one body of hand) are interconnected into one kinematic chain (Figure 1). The distal end of arm is connected with the forearm by a joint (3) with two degrees of freedom. The forearm is connected with the hand (10) by a joint (5) with two degrees of freedom. Each body of the model has a defined mass, moment of inertia, center of gravity and geometry (Table 1). Properties of the bodies were calculated based on data obtained from the Generator of Body Data

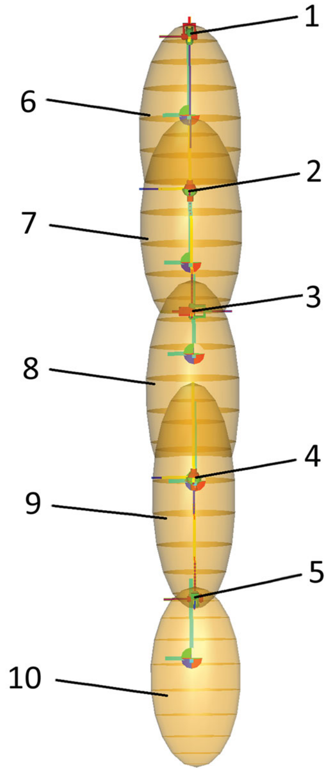

Figure 1. Upper extremity kinematic chain.
Note: 1 = shoulder joint; 2 = arm fracture point (joint that connects two parts of the arm); 3 = elbow joint; 4 = forearm fracture point (joint that connects two parts of the arm); 5 = wrist joint; 6 = upper part of the arm; 7 = lower part of the arm body; 8 = upper part of the forearm body; 9 = lower part of the forearm body; 10 = hand body.

(GEBOD).[8] The geometry of the model was created using ellipsoids.

3.1. Fracture model

As already shown, both the arm and the forearm consist of two rigid bodies each connected by joints with three degrees of freedom. These joints are so-called fracture points (Figure 1, 2 and 4). For each one of them, limiting force and bending moment were defined. These limits were calculated on the basis of fracture tolerance data taken from the literature (Table 2) as described in Section 3.2.

Simulation of a fracture works as follows. If no force/bending moment is applied to the considered extremity segment (arm or forearm) or the force/bending moment is lower than the restraint limit value, then the segment remains in a neutral position (Figure 2(a)). But when the force/bending moment applied to the segment is higher than the restraint limit value, then the connection breaks (Figure 2(b)). There are two fracture points in the model. The first fracture point is located at half the length of the arm and is used to simulate a fracture of the humerus bone. The second fracture point is located at half the length of the forearm and simulates a fracture of the radius and ulna bones.

Table 1. Properties of the arm and forearm calculated for the 50th percentile male based on data obtained from the Generator of Body Data (GEBOD).[8]

Body segment	Mass (kg)	Inertia (I_x, I_y, I_z) (kg m^2)	Center of gravity (x, y, z) (m)	Ellipsoid semi-axes x, y, z (m)
Upper arm (Figure 1, 6)	0.9488	0.0016 0.0021 0.0011	0.0009 − 0.0025 − 0.0741	0.0482 0.0448 0.1041
Lower arm (Figure 1, 7)	0.9488	0.0016 0.0021 0.0011	0 0 − 0.0693	0.0482 0.0442 0.1041
Upper forearm (Figure 1, 8)	0.6669	0.0016 0.0016 0.0007	− 0.0013 − 0.0017 − 0.0400	0.0442 0.0402 0.1078
Lower forearm (Figure 1, 9)	0.6669	0.0016 0.0016 0.0007	− 0.0013 − 0.0017 − 0.0035	0.0392 0.0372 0.1078
Hand (Figure 1, 10)	0.5010	0.0011 0.0013 0.0004	0.0037 0.0018 0.0018	0.0440 0.0140 0.1090

Table 2. Values of the arm and forearm fracture forces and bending moments normalized for the 50th percentile male.

50th percentile male	Mass m_{ref} (kg)	Length L_{ref} (cm)	Normalized fracture force (N), direction: lateral/medial	Normalized fracture bending moment (Nm), direction: lateral/medial	Normalized fracture force (N), direction: anterior/posterior	Normalized fracture bending moment (Nm), direction: anterior/posterior
Whole body	78.70	177.30				
Arm	1.90	33.40	2560[a]	245[a]	2560	245
Forearm	1.34	30.60	1715	120	1805[b]	80[b]

Note: [a]Kirkisch et al. [9] used the same scaling method but they used length and mass of the arm from a different source; the result they obtained was force $F = 2500$ N and moment $M = 230$ Nm.
[b]These values are calculated for the forearm in supinated position. Duma et al. [13] noticed that the forearm is about 21% stronger in the pronated position than in the supinated position.

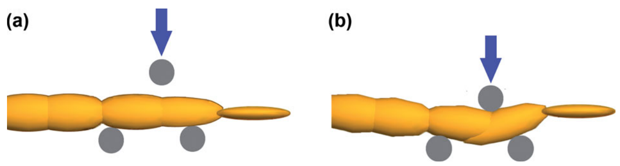

Figure 2. Force applied to the upper extremity: (a) before applying; and (b) during applying (limit value exceeded).

3.2. *Fracture tolerance data*

Fracture tolerance data include extreme values of the forces and bending moments which can cause fractures of the bone (hereinafter referred to as fracture forces and fracture bending moments). Values of the fracture forces and bending moments were taken from the literature and recalculated using a normalization and scaling algorithm described in the following.

Fracture forces and bending moments of the arm were calculated on the basis of the results of Kirkish et al.'s [9] experiment. Kirkish et al. conducted three-point bending tests on an Instron (USA) testing machine to determine forces and bending moments, which cause humerus fracture. Tests were performed with 26 human arms (specimens) taken from male cadavers and one female cadaver. The cadavers ranged in age from 44 to 75 years. The arms were disarticulated from the body and from the forearm. The soft tissue was not removed except for the ends of the humerus. At both ends of the bone, epoxy blocks were molded and the prepared upper arm was placed in the testing machine. Epoxy blocks were placed on 12.5-mm diameter rollers at each specimen end. Load was applied at half the arm length until fracture. Tests were carried out in two loading directions: posterior and medial. As noted by Kirkish et al. on the basis of the analyses of the results, there is no significant difference between posterior and medial loading directions and in both cases the forces and moments were similar.

Fracture forces and bending moments of the forearm were calculated on the basis of the results of Pintar and Yoganandan's [10] and Begeman and Pratima's [11] experiments. In both cases experiments were conducted on the whole forearm, with the radius and ulna bones together.

Pintar and Yoganandan conducted dynamic three-point bending failure tests on 12 female and 18 male forearms. Specimens were taken from 30 cadavers ranging in age from 41 to 89 years. Specimens were placed on two supports in a supinated position and force was applied at half the forearm length. An impactor was loading the forearm in a posterior direction. Begeman and Pratima conducted three-point bending dynamic and quasi-static tests on 20 human forearms. All specimens were taken from male cadavers ranging in age from 50 to 73 years. Tests were conducted in two configurations. In the first, loading was applied in a medial direction (force applied first to the radius bone). In the second configuration, loading was applied in a lateral direction (force applied first to the ulna bone). In both cases, specimens were placed on supports in a supinated position. The differences between the loading directions were negligible.

In all of these experiments specimens were taken from cadavers of different age, gender, dimensions, masses, etc. These differences influence the fracture forces and bending moments. In order to obtain values for a particular group of people, experimental results must be normalized (scaled). During the normalization process it is possible to consider the weight and height of the human (age, gender or state of health are not considered in the commonly used normalization methods). This means that it is possible to normalize data, e.g., for the 5th, 50th or 95th percentile male or female. In our case, data were normalized for the 50th percentile male because that group of people is most often injured in accidents at work. For this task normalization, the method described by Kirkish et al. [9] was used. This method was originally developed by Mertz [12].

The normalization was conducted in order to obtain data of fracture forces and bending moments for the 50th percentile male. On the basis of lengths and weights of the arm and forearm, the scaling factors were calculated using Equations (1), (2) and (3):

$$\lambda_z = \frac{L_{\text{norm}}}{L_{\text{ref}}}, \tag{1}$$

$$R_{\text{m}} = \frac{m_{\text{norm}}}{m_{\text{ref}}}, \tag{2}$$

$$\lambda_x = \lambda_y = \sqrt{\frac{R_{\text{m}}}{\lambda_z}}, \tag{3}$$

where λ_z = scale factor along the z axis (longitudinal axis) of the extremity segment (arm or forearm), L_{norm} = length of the extremity segment (arm or forearm) of the 50th percentile male, L_{ref} = reference length of the extremity segment taken from the experiment, R_{m} = scale factor is ratio of reference mass of the extremity segment to the 50th percentile male extremity segment, m_{norm} = mass of the extremity segment of the 50th percentile male, m_{ref} = reference mass of the extremity segment taken from the experiment and λ_x, λ_y = scale factors along the x and y axes (i.e., axes parallel to the cross-section of the extremity segment) of the extremity segment (arm or forearm).

Table 3. Comparison of the fracture forces and bending moments normalized for the 5th percentile female.

5th percentile female	Our results	Kirkisch et al.'s results [9]	Duma et al.'s results [13]
Arm			
F (N)	1713.31	1700	–
M (Nm)	134.14	130	128 ± 19
Forearm			
F (N)	1526.08	–	–
M (Nm)	62.19	–	58 ± 12

Note: F = force; M = moment.

Lengths and weights of these segments of the 50th percentile male were taken from GEBOD.[8]

The second stage consisted of calculating the normalized fracture force and bending moment (Equations (4) and (5)):

$$F_{\text{norm}} = \lambda_x \lambda_y F_{\text{ref}}\ (N), \tag{4}$$

$$M_{\text{norm}} = \lambda_x^2 \lambda_y M_{\text{ref}}\ (\text{Nm}), \tag{5}$$

where F_{norm} = normalized fracture force, λ_x, λ_y = scale factors along the x and y axes (i.e., axes parallel to the cross-section of the extremity segment) of the extremity segment (arm or forearm), F_{ref} = reference fracture force taken from the experiment, M_{norm} = normalized fracture bending moment and M_{ref} = reference fracture bending moment taken from the experiment.

Results of the calculations are presented in Table 2. These results were used as values of fracture forces and bending moments in the model of the human upper extremity.

Fracture forces and bending moments can also be scaled for a different percentile of the human population. This can be done on the basis of the data from Table 2 and Equations (1)–(5). In order to verify this procedure, normalized fracture forces and bending moments were calculated for the 5th percentile female and compared with appropriate data obtained from the literature (Table 3). It should be note that in this case, length (L_{norm}) and mass (m_{norm}) from Equations (1) and (2) had to be obtained for the 5th percentile female instead of the 50th percentile male.

The 5th percentile female was calculated as follows:

- Data on masses and lengths of the 5th percentile female arm and forearm were taken from GEBOD:

Arm:	Forearm:
L_{norm} (arm) = 30.48 cm	L_{norm} (forearm) = 29.2 cm
m_{norm} (arm) = 1.159 kg	m_{norm} (forearm) = 1.08 kg

- Reference masses and lengths of the arm and forearm were taken from Table 2:

Arm:	Forearm:
L_{ref} (arm) = 33.4 cm	L_{ref} (forearm) = 30.6 cm
m_{ref} (arm) = 1.9 kg	m_{ref} (forearm) = 1.34 kg

- The scaling factors were calculated based on Equations (1), (2) and (3):

Arm:	Forearm:
λ_z (arm) = 0.912	λ_z (forearm) = 0.954
R_{m} (arm) = 0.611	R_{m} (forearm) = 0.807
λ_x (arm) = 0.818	λ_x (forearm) = 0.919
λ_y (arm) = 0.818	λ_y (forearm) = 0.919

- Reference fracture forces and bending moments of the arm and forearm were taken from Table 2:

Arm:	Forearm:
F_{ref} (arm) = 2560 N	F_{ref} (forearm) = 1805 N
M_{ref} (arm) = 245 Nm	M_{ref} (forearm) = 80 Nm

- Normalized fracture forces and bending moments were calculated for the 5th percentile female based on Equations (4) and (5):

Arm:	Forearm:
Normalized values for lateral, medial, anterior and posterior directions:	Normalized values for anterior and posterior directions:
F_{norm} (arm) = 1713.31 N	F_{norm} (forearm) = 1526.08 N
M_{norm} (arm) = 134.14 Nm	M_{norm} (forearm) = 62.19 Nm

3.3. *Sample use of the model*

A simple example of a reconstruction of a real-life accident at work will illustrate the application of the model of the upper extremity. The model was implemented into the pedestrian human model (the human model from the MADYMO database).[14]

The accident happened in a warehouse. The worker was standing on the footrest of a container, he was holding the flap (Figure 3). Suddenly he slipped from the footrest and the flap pinned his left upper extremity to the container. This resulted in fractures of both bones of the left forearm. A numerical reconstruction is rarely used in such accidents because the accident is very simple and the course of the accident is clear. But in this case it was very important to choose as simple a case as possible (with upper extremity injuries only) in order to verify the possibility of simulating a fracture of the upper extremity.

A computer model of the accident site was created. It consisted of the container with a moveable flap and the human model. The simulation was run for 1 s. The animation output is shown in Figure 4.

Figure 3. Simulation set-up.

Virtual sensors measured bending moments in forearms and arms. Table 4 presents results only for the left upper extremity (the flap did not hit the right upper extremity).

A comparison of the simulation results with the values for the injury criteria for the arm and forearm shows that the bending moment for the left forearm (direction: anterior/posterior) exceeded the critical values, which means a high probability of a fracture of forearm bones. According to the description of the accident, that is exactly what happened to the worker.

4. Discussion and conclusion

Knowledge of the mechanical properties of the human body is very important in developing numerical models of human and crash test dummies. For the purposes of developing such models a number of experimental tests on cadavers are conducted. Most models are developed for the purposes of the automotive and aerospace industry. These models make it possible to evaluate injuries of the head, neck, chest and lower extremities. Most human or dummy models have a simplified model of the upper extremity. These models do not make it possible to evaluate injuries of the upper extremities. However, in some cases it is necessary to evaluate them, e.g., in the case of a biomechanical analysis of injuries caused by deploying airbags or reconstructions of accidents related to work and sport. There are very few numerical models of the upper extremity, with possibilities of evaluating injuries, described in the literature (only two models have been found in the available

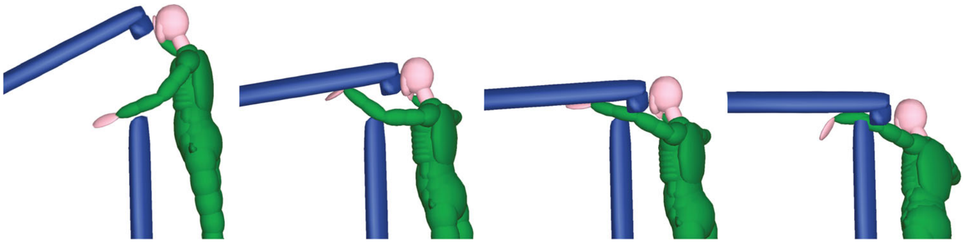

Figure 4. Final phase of simulation in MADYMO R7.5.

Table 4. Values of injury criteria for left upper extremity.

Injury criterion	Value	Critical value	Description
Left arm			
Bending moment (Nm), direction: lateral/medial	15.65	245	No fracture
Bending moment (Nm), direction: anterior/posterior	8.69	245	No fracture
Left forearm			
Bending moment (Nm), direction: lateral/medial	57.16	120	No fracture
Bending moment (Nm), direction: anterior/posterior	121.84	80	High probability of fracture of forearm bones

literature). Both models [15,16] had been developed with use of the FE method. Moreover, both models are capable of predicting kinematics and potential injury when interacting with a deploying airbag. The former model [15] can be integrated with FE and rigid body dummy models available in PAM-CRASH software (currently Virtual Performance Solution [17]). The model had been developed on the basis of experimental tests on cadavers. A similar FE upper extremity model [16] had been developed for use with the MADYMO human model. Both models consist of bones modeled as a shell, soft tissues modeled as a solid elements and skin modeled as a shell. Such a solution gives a more accurate geometry representation compared with the MB model we developed and more accurate simulation results at the same time (FE models allow a detailed analysis, which takes into account, e.g., the exact location of a fracture). However, these models have a major disadvantage when they are used for reconstructing accidents at work. FE simulation is more accurate and provides more complex results; however, it is much slower than MB. Typical FE simulations in automotive accident reconstructions take less than 250 ms; however, it takes a few or even a few dozen hours of CPU time to complete calculations. Reconstructions of accidents at work take much longer, up to 10 s. The MB method can produce results in more reasonable time. At the same time, MB is sufficiently accurate for accident simulations. Therefore, a numerical model of the upper extremity was developed using the MB method.

In the current version of the model of the upper extremity, the values of fracture forces and bending moments have been defined. The main advantage of the model is an opportunity to evaluate quickly whether a fracture of the bone is possible. The disadvantage of such a solution is that it is impossible to assess the probability of bone fracture. However, it is possible to develop an injury risk function for the 50th percentile male using available experimental data, e.g., on the basis of Duma et al.'s studies.[13] An advanced user can easily modify the model by locking the fracture points. After modification, the model will measure forces and bending moments without fracture modeling. In order to evaluate the probability of a bone fracture, the user will be able to compare measured values with the reference values from the developed injury risk function.

The model is intended to be used with the MADYMO pedestrian human model; however, it is possible to connect it to other human models. The model of the upper extremity and fracture simulation data were created to reflect the 50th percentile male, but the model can be scaled for a different percentile of the human population. For this task, data obtained, e.g., from GEBOD can be used.

The model was developed for the purposes of reconstructing accidents at work but it can also be applicable in other biomechanical analyses. As already mentioned, this model is used with the MADYMO pedestrian human model. A model of a pedestrian with an upper extremity can be used in analyzing car-to-pedestrian accidents, motorcycle or bicycle accidents, or sport accidents.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

This paper is based on the results of a research task carried out within the scope of the second stage of the National Programme 'Improvement of safety and working conditions' partly supported in 2011–2013 – within the scope of research and development – by the Ministry of Science and Higher Education/National Centre for Research and Development. The Central Institute for Labour Protection – National Research Institute (CIOP-PIB) was the Programme's main co-ordinator.

References

[1] Ibitoye A, Hamouda A, Wong S, et al. Simulation of motorcyclist's kinematics during impact with W-Beam guardrail. Advances in Engineering Software. 2006;37:56–61. doi:10.1016/j.advengsoft.2004.12.002

[2] TASS International. [cited 2014 Feb 3]. Available from: http://www.tassinternational.com/madymo.

[3] Livermore Software Technology Corp. [cited 2014 Feb 3]. Available from: http://www.lstc.com/products/ls-dyna.

[4] Konarzewski K, Matyjewski M, Rzymkowski C. Wybrane zagadnienia biomechaniki zderzeń [Selected issues in impact biomechanics]. In: Nałęcz M, Będziński R, Kędzior K, et al., editors. Biomechanika i inżynieria rehabilitacyjna [Biomechanics and rehabilitation engineering]. Warszawa: Exit; 2000. p. 747–797.

[5] Milanowicz M, Budziszewski P. Numerical reconstruction of the real-life fatal accident at work: a case study. Lecture Notes in Computer Science. Duffy VG, editor. DHM/HCII 2013, Part II. 2013;8026:101–110. doi:10.1007/978-3-642-39182-8_12

[6] Milanowicz M, Budziszewski P. Wykorzystanie komputerowego modelu człowieka do rekonstrukcji wypadków przy pracy [The use of computer human model for the reconstruction of accidents at work]. Mechanik. 2011;7:567–574.

[7] TASS International. MADYMO theory manual [cited 2014 Feb 3]. Available from: http://www.tassinternational.com/software-support.

[8] Cheng H, Obergefell L, Rizer A. Generator of body data (GEBOD) manual. Report No. AL/CF-TR-1994-0051; 1994.

[9] Kirkish S, Begeman P, Paravasthu N. Proposed provisional reference values for the humerus for evaluation of injury potential. SAE Technical Paper No. 962416. Warrendale (PA): Society of Automotive Engineers; 1996.

[10] Pintar A, Yoganandan N. Response and tolerance of the human forearm to impact loading. SAE Technical Paper No. 983149. Warrendale (PA): Society of Automotive Engineers; 1998.

[11] Begeman P, Pratima K. Bending strength of the human cadaveric forearm due to lateral loads. SAE Technical Paper No. 99SC24. Warrendale (PA): Society of Automotive Engineers; 1999.

[12] Mertz H. A Procedure for normalizing lmpact response data. SAE Technical Paper No. 840884. Warrendale (PA): Society of Automotive Engineers; 1984.

[13] Duma S, Boggess B, Bass C. et al. Injury risk functions for the 5th percentile female upper extremity. SAE Technical Paper No. 2003-01-0166. Warrendale (PA): Society of Automotive Engineers; 2003.

[14] TASS International. MADYMO human models manual [cited 2014 Feb 3]. Available from: http://www.tassinternational.com/software-support.

[15] Palaniappan Jr P, Wipasuramonton P, Begeman P, et al. A three-dimensional finite element model of the human arm. SAE Technical Paper No. 99SC25. Warrendale (PA): Society of Automotive Engineers; 1999.

[16] van Rooij L, Bours R, van Hoof J, et al. The development, validation and application of a finite element upper extremity model subjected to air bag loading. Stapp Car Crash Journal. 2003;47(10):55–78.

[17] ESI-Group. [cited 2014 Feb 3]. Available from: http://www.esi-group.com.