Acta of Bioengineering and Biomechanics
Vol. 18, No. 1, 2016

Original paper
DOI: 10.5277/ABB-00214-2014-03

Experimental study of the loss of balance process before falling from a height

MARCIN MILANOWICZ[1]*, PAWEŁ BUDZISZEWSKI[1], KRZYSZTOF KĘDZIOR[2]

[1] Central Institute for Labour Protection – National Research Institute, Department of Safety Engineering, Warsaw, Poland.
[2] Central Institute for Labour Protection – National Research Institute, Department of Ergonomics, Warsaw, Poland.

Due to a high number of accidents at work, actions have been undertaken to apply numerical simulation for reconstruction of their course in time. First works on this issue showed that the numerical human body model developed in the Central Institute for Labour Protection – National Research Institute (CIOP-PIB) should be adjusted to the nature of specific accidents. Hence, a need arose to adjust the model for reconstruction and simulation of falls from heights. The adjustment involved supplementing the model with functions enabling taking into account movements of a person at the time of losing their balance, as such movements influence the course of a fall. For this purpose, a study with participation of volunteers has been conducted to define these movements. Within the framework of the study, reactions (body movements) of the subjects at the time of losing their balance were recorded. The works resulted in obtaining parameters that, after an analysis, may be used as initial conditions for a numerical model of the human body.

Key words: *loss of balance, falls from height, 3D motion analysis, virtual reality*

1. Introduction

According to data provided by the Central Statistical Office of Poland (GUS), every year the number of persons injured in accidents at work in Poland remains at the level of about 90,000 [9]. The most numerous group of such incidents includes slipping, tripping and falling from heights. They constitute an average of 25–30% of all accidents at work. The most common incidents resulting in falling from a height are: loss of balance caused by instability of a material agent (e.g., a ladder), slipping, tripping, climbing posts, scaffolds and other structures and loss of balance during works on a roof edge [7].

Due to the large number of accidents at work, for several years the Central Institute for Labour Protection – National Research Institute (CIOP-PIB, Warsaw, Poland) has been conducting works on application of numerical methods for reconstruction of accidents related to mechanical hazards [17]. The basic assumption for application of numerical methods for reconstruction of accidents is the need to recreate a real accident situation by means of a numerical simulation. Numerical modelling of physical phenomena enables recreation of a real or close to real course of an accident, its reasons and outcomes, based on the laws of physics. Simulation of accidents at CIOP-PIB uses a MADYMO numerical model of the human body by TASS [14], which is an accurate reflection of human kinetics and enables an assessment of injuries. However, reconstruction works have shown some limitations to the existing model, preventing a complete and reliable analysis of some types of accidents, among others, accidents related to falling [16]. Currently, numerical reconstruction and simulation of accidents related to falling uses passive numerical models of the human body, i.e., models maintaining general human

* Corresponding author: Marcin Milanowicz, Central Institute for Labour Protection – National Research Institute, Department of Safety Engineering, ul. Czerniakowska 16, 00-701 Warsaw, Poland. Phone number: +48 22 623 46 66, fax: +48 22 623 3693, e-mail: marmi@ciop.pl

Received: October 29th, 2014
Accepted for publication: April 30th, 2015

kinetics but not taking into account movements of the body resulting from muscle tension. This means that during simulation of a fall, the numerical model describes an inert fall of the human body to the ground. Meanwhile, a crucial factor influencing the trajectory of a fall is the reaction of a person at the time of losing their balance. At that time, the falling person determines the initial conditions constituting the input data to begin a numerical simulation of a fall. In order to include human reactions into a reconstruction of a fall from a height, the numerical model needs to be supplemented with functions describing the initial movement of the human body at the time of loss of balance, e.g., functions including information about values of angles at a given moment in time, as measured for individual human joints. Consequently, simulation of a fall will be more accurate. In order to obtain functions describing the initial conditions, it is necessary to conduct studies with participation of volunteers, in order to record movements of a person at the time of losing their balance. Conducting such studies in real conditions, i.e., initiating a fall of a subject from a considerable height, is impossible for two reasons: firstly, due to exposure of the subject to a real risk of damaging their health or losing their life; secondly, due to the fact that the course of the test should be recorded with special equipment, very difficult to install in such conditions. As a result, studies need to be conducted within a limited scope, i.e., in laboratory conditions at the ground level or at a very small height above the ground. Unfortunately, a laboratory does not resemble conditions of working at heights, so initiating loss of balance in such conditions could produce results inconsistent with the reality. A solution to this problem is the modern technology of immersion virtual reality (VR). The VR technology involves isolation of a test subject from visual and sound stimuli of the real environment, which are replaced by an image and sound of a presented simulated world. This is executed with the use of a device called HMD (Head Mounted Display), with a structure resembling goggles with a display mounted in front of each of the eyes. The rest of the field of view is covered, isolating a person from visual sensations of the real world. Additionally, sensors tracing movement of the head and hands installed in an HMD and held in hands enable looking around and moving across the virtual environment in a natural way executed with movements of the human body. Due to application of this technology, it is possible to create a virtual construction site and "transfer" a subject to scaffold so that they have an impression of working at a very great height.

2. Materials and methods

2.1. Study population

30 males at the age of 22.90 ± 1.94, of a weight of 79.70 ± 8.93 kg and a height of 181.83 ± 6.94 cm volunteered in the study. The subjects had no previous experience in working at heights due to the fact that most accidents happen among young and inexperienced workers. Before the test, every subject was given general information about the study and signed a consent to perform the tests. Since the element of surprise, in the form of an unexpected loss of balance, played a crucial role in the study, the subjects were not informed about the actual purpose of the study itself. However, they knew that during the study a situation might (while not necessarily will) occur that could result in loss of balance.

2.2. Test rig

In order to conduct the study, a test rig was designed and made to enable simulation of working at a height. The main element of the rig was a computer program resembling a computer game with a visualisation of a construction site with scaffolds and construction equipment. The program was displayed on an HMD screen, so that after putting on the VR equipment, the test subject had an impression of being on a scaffold 20 m above the ground. Design of the program enabled execution of various construction works. Looking around the environment was made possible with the use of two Razer Hydra [22] sensors recording location in space. One sensor was mounted on the HMD and the other was held in the subject's right hand. The hand-held controller was equipped with a small joystick used to control a rope winch in the virtual work environment.

Another component of the rig was a special device used to unbalance the subjects. The device is made of aluminium alloys and is composed of two main parts: a fixed base and a swivelling platform, linked to the base with a bearing mounting. The base consists of a frame of the dimensions of 800 × 1,200 mm and an extension arm; the platform is made of a metal frame of the dimensions of 500 × 800 mm to which an anti-skid mat of the same dimensions has been fixed. Thanks to the application of the extension arm, it is possible to adjust the height of the platform within the range of 250–375 mm. In the initial condition the platform is positioned in parallel to the ground and is locked in this position with the use of a SCOT EL 1200SL electromagnetic jumper, locking the platform to the base with a force of ca. 5.4 kN. The electromagnetic jumper works correctly (i.e., closes both elements of the platform) after applying a voltage of 24 V. Cutting the voltage off results in switching off the electromagnet, enabling the platform to tilt. The electromagnetic jumper was connected to the computer with an Arduino Uno microcontroller. The jumper was controlled from the level of the computer program (Fig. 1).

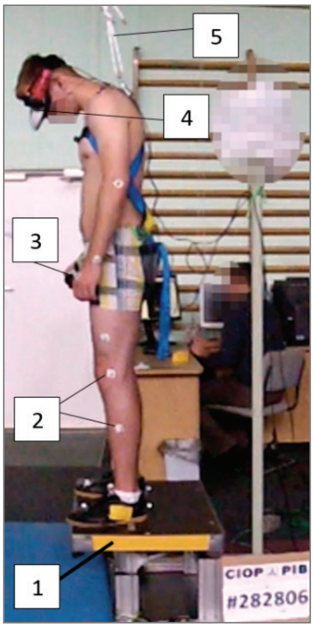

Fig. 1. Test rig configuration. 1 – platform, 2 – VICON markers, 3 – Razer Hydra controller, 4 – HMD, 5 – belay rope

2.3. Procedure of the study

The purpose of the study was to obtain data describing movements resulting from human muscle tension due to loss of balance. The study was conducted in the Central Research Laboratory of the Academy of Physical Education in Warsaw. A permit to conduct the study was issued by the Senate Committee for Ethics in Scientific Research of the Józef Piłsudski Academy of Physical Education in Warsaw (No SKE 01-05/2013 of 8.03.2013).

Before the test, test rig set up was introduced to the subjects and the course of study was explained. During the test, a subject stood on the edge of the platform with an HMD on his head. He held a joystick controller in his hand. The HMD displayed an image of a virtual work environment, so the subject had an impression of standing on the edge of scaffold, about 20 m above the ground. The subject's task was to control a virtual rope winch. He looked down and used the joystick to control an object placed at the end of a rope. Due to the design of the task, initially each subject of the study stood in a similar position and at the same place. At the same time, this task required focus, so a subject did not expect that an incident might occur any time and result in falling from a height.

Initially the platform was locked and constituted a stable platform for the subject. The platform was placed 390 mm above the floor. At a moment unexpected for the subject an incident was initiated involving sliding off the platform that lock was released and the platform tilted by an angle of almost 90 degrees. At that time, the subject lost his balance and fell on a prepared mattress. In order to prevent tripping, every subject was wearing a harness for work at heights attached to a belay rope.

2.4. Measuring instruments

Movements of the subjects at the time of losing their balance were captured by means of VICONMx [27], a system for 3D motion analysis. Body movement was recorded with 9 digital cameras filming with the frequency of 100 Hz. The cameras recorded location of 34 light reflexive markers stuck on various spots of the human body. Location of the markers complied with the Full Body PlugInGait (SACR) protocol. It is a protocol included in the VICON system and precisely describing the spots on the human body where markers should be placed.

The main result of the filming was a 3D recording of location of 34 markers in the VICON coordinate system. VICON has its own software to control the cameras. Before recording a test, the program creates a virtual model of the subject's skeleton. For this purpose, the person controlling the system selects each marker individually and assigns it to a respective body part. This way every marker placed on the subject's body is assigned to a respective body part.

Recorded marker locations are used by the system to calculate angles in individual joints. During the study angles in 14 human joints were recorded (neck, shoulders, elbows, wrists, spine, hips, knees, ankles) and 3 angles measured between the head, chest, pelvis and a system of inertial coordinates related to the laboratory. The beginning of the coordinate system (0,0,0) is located on the floor, at a distance of X = 342.5 mm from the centre of the platform, Y = 1.280 mm from the platform rotation

Fig. 2. Test course filmed using GoPro Hero2 camera for one of the subjects

axis and Z = 390 mm from the surface on which the subject stands (see Fig. 1).

Additionally, the study was filmed using two GoPro Hero2 cameras [10], recording image with the frequency of 120 fps. Camera placement enabled filming of the experiment from the side (Fig. 2) and from the front.

3. Results

Results of the study have a form of trajectories for individual parts of the subject's body. On the basis of the recorded trajectories, the following parameters were calculated:

- positions, speeds and accelerations of the body centre of gravity in 3D,
- values of angles, speeds and angle accelerations for selected body parts and most of human joints.

Within the framework of the study, 30 reactions (body movements) of the subjects at the time of losing their balance were recorded. Completion of the test recording for every subject was determined individually and was from 40 to 65 ms from the moment of releasing the lock. Finishing recording after such a short time was necessary, because the belay rope stretched, causing a change in trajectories of the fall (the human body was hampered and pulled to the front). Different timing for every subject results from the fact that the rope stretched at a different moment for every subject.

The histograms show values of angles measured in the selected human joints at the finishing moment of a test recording. The most frequent angle values for individual measurement points are:

– The angle between the chest and the pelvis in the lumbar section, in the sagittal plane, the X-axis (flexion – positive values, extension – negative values); 15÷30 degrees (Fig. 3);
– The angle in the lumbar section, in the frontal plane, the Y-axis (lateral bending to the right – positive values, lateral bending to the left – negative values): –4÷4 degrees (Fig. 3);
– The angle in the lumbar section, in the transverse plane, the Z-axis (rotation to the right – positive values, rotation to the left – negative values): –5÷5 degrees (Fig. 3);
– Angles in the hip joints, in the sagittal plane (flexion – positive values): 0÷30 degrees (left), 5÷25 degrees (right) (Fig. 4);

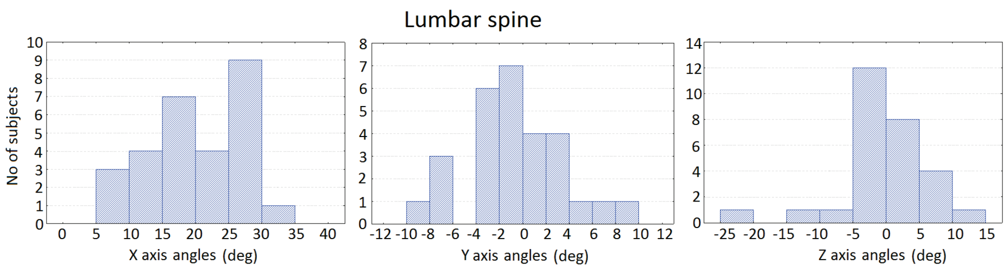

Fig. 3. Angle values at the finishing moment of a test measurement recorded in the lumbar spine section

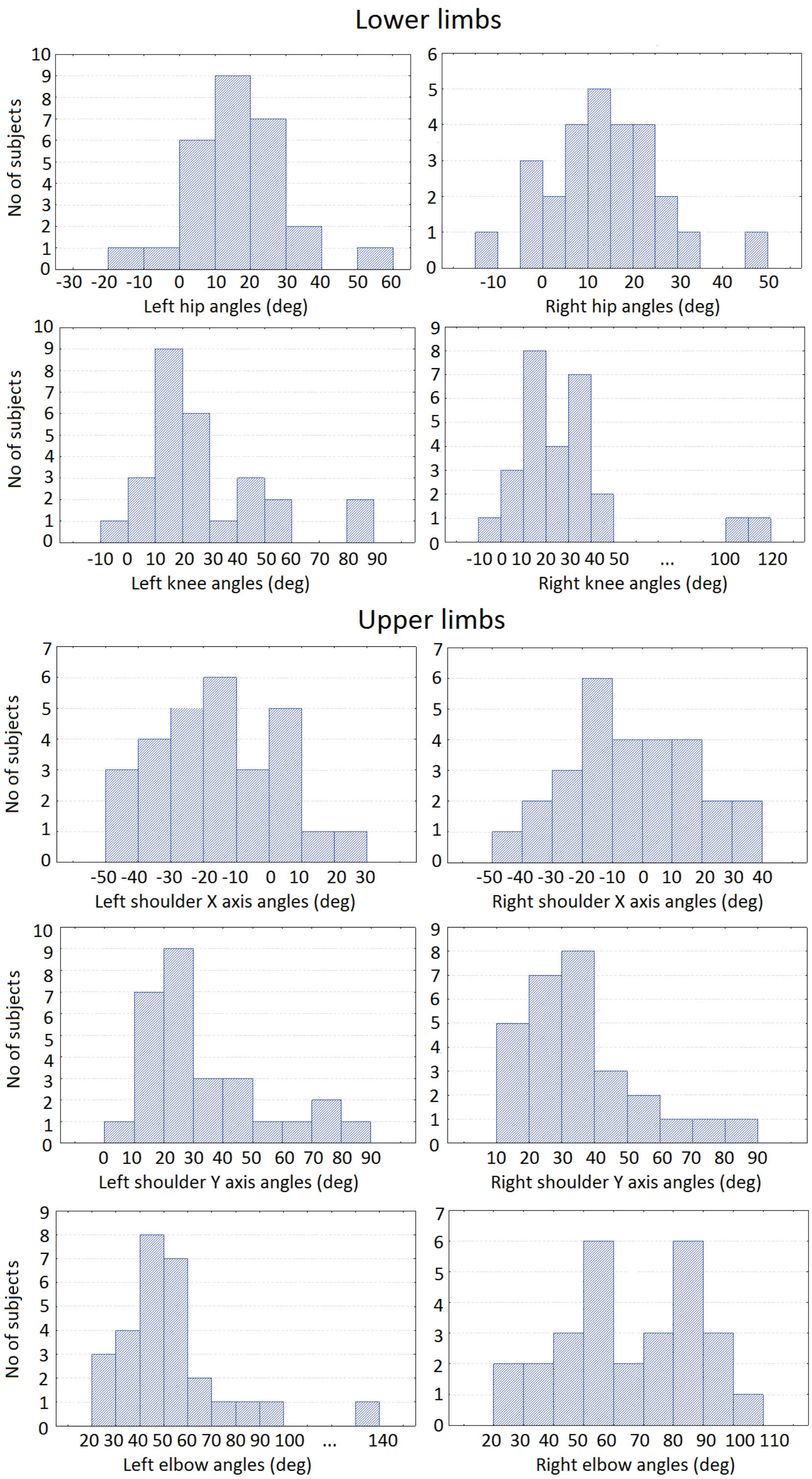

Fig. 4. Angle values at the finishing moment of a test measurement recorded in the joints of the upper and lower limbs

- Angles in the knee joints, in the sagittal plane (flexion – positive values): 10÷30 degrees (left), 10÷40 degrees (right) (Fig. 4);
- Angles in the shoulder joints, in the sagittal plane, the *X*-axis (flexion – positive values): –30÷10 degrees (left), –20÷20 degrees (right) (Fig. 4);
- Angles in the shoulder joints, in the frontal plane, the *Y*-axis (abduction – positive values): 10÷30 degrees (left), 10÷40 degrees (right) (Fig. 4);
- Angles in the elbow joints, in the sagittal plane (flexion – positive values): 40÷60 degrees (left), 50÷90 degrees (right) (Fig. 4).

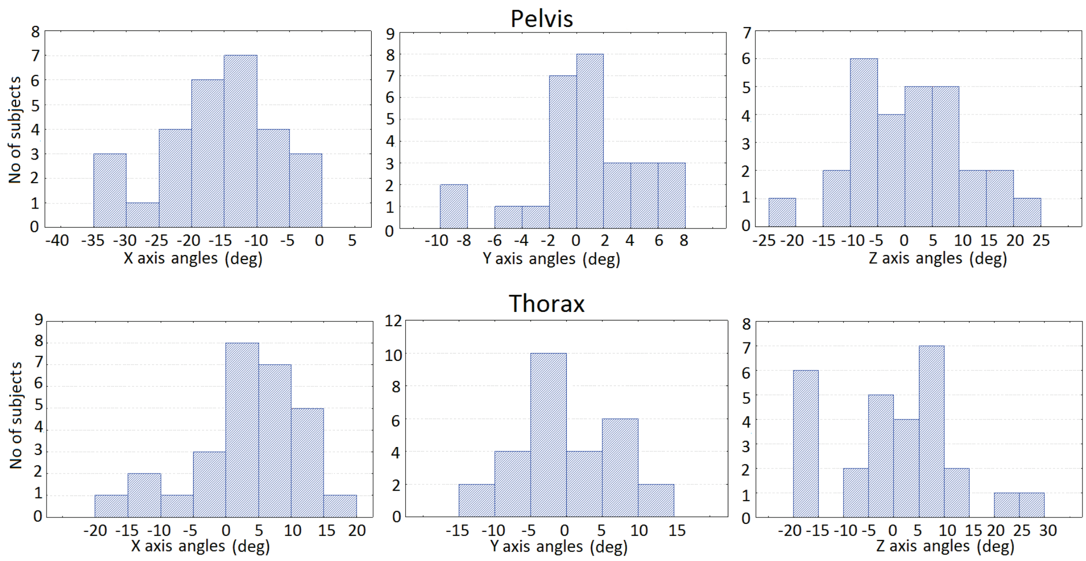

Fig. 5. Angle values at the finishing moment of a test measurement recorded for the pelvis and chest

Chest and pelvis rotation in relation to the coordinate system of the laboratory:

– The chest angle in the sagittal plane, around the *X*-axis (flexion – positive values, extension – negative values): 0÷15 degrees (Fig. 5);
– The chest angle in the frontal plane, around the *Y*-axis (rotation to the right – positive values, rotation to the left – negative values): –5÷5 degrees (Fig. 5);
– The chest angle in the transverse plane, around the *Z*-axis (rotation to the right – positive values, rotation to the left – negative values): –20÷–15 and –5÷10 degrees (Fig. 5);
– The pelvis angle in the sagittal plane: –25÷–5 degrees (Fig. 5);
– The pelvis angle in the frontal plane: –2÷4 degrees (Fig. 5);
– The pelvis angle in the transverse plane: –10÷10 degrees (Fig. 5).

Results of 28 subjects underwent further analysis; results of two subjects were eliminated due to technical mistakes (measurement of some angles had not been recorded). Works determined the phases of human reactions that occurred subsequently during the test and parameters influencing the course of a human reaction at the time of loss of balance were investigated.

The moment of loss of balance was divided into three phases (Fig. 6):

Phase 1: The time from releasing the platform lock to the moment of first human reaction. In this time, plantarflexion of the feet in the ankle joint begins, due to a change in the platform angle. This phase lasts an average of 10.85 ms.

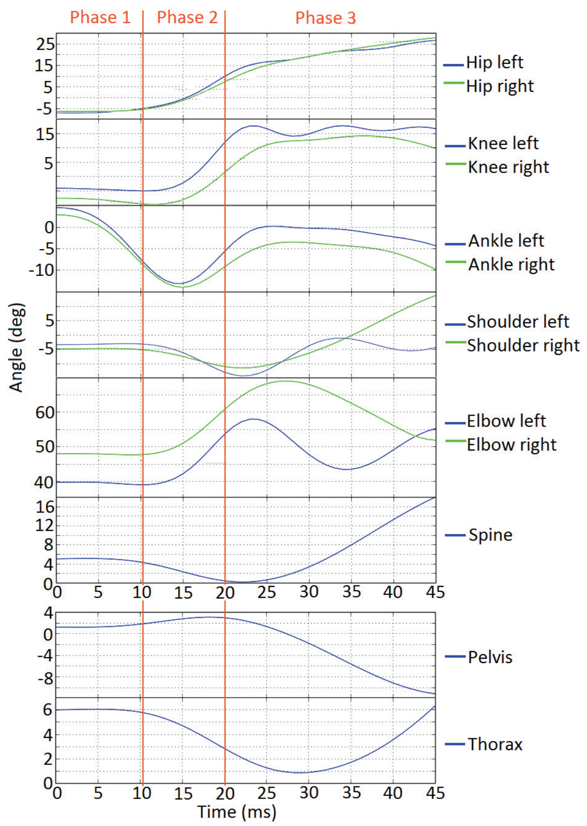

Fig. 6. Example of a recorded angle transition for one of the subjects, showing phases of human reaction at the time of loss of balance

Phase 2: Human reaction to tilting of the platform. The earliest noticeable human reaction to a sudden change of the platform position is the sudden flexion of the lower limbs in hip joints. At the same time, or several milliseconds later, a reaction is visible in the knee joints (flexion) and in the upper limb joints (flexion with simultaneous abduction in the shoulder joints and flexion in the elbow joints). In most cases, the reaction in the ankle joints (dosiflexion) is the last to occur at this phase – on average 6.25 ms after the initial reaction in the hip joints. Duration of this phase is an average of 12.55 ms.

Phase 3: Flexion in the lumbar spine section. A movement measured in the lumbar spine section is the last to be noticed, hence this motion has been classified as the next phase. Flexion of the upper body in relation to the pelvis is visible in this phase. This reaction is frequently accompanied by sudden extension and re-flexion of the upper and lower limbs in the joints or a slowdown of flexion in these joints. This phase ends with the end of the measurement. It lasts an average of 24.5 ms.

All 3 phases result in pelvis deflection in relation to the coordinate system of the laboratory. However, the chest initially deflects backwards, but then, due to flexion in the lumbar spine section, forward deflection occurs. The entire body, however, shows a tendency to deflect. What is more, some of the subjects additionally leaned and rotated to the right or left.

Within the framework of the analysis, parameters influencing the body angular position and velocity at the finishing moment of the test measurement were investigated. The following parameters were selected: initial value of the chest, pelvis, hip angles, maximum rotational speed of the lumbar spine and chest angles at the finishing moment of a test recording. The Shapiro–Wilk test showed that data within these parameters were normally distributed. A Pearson product-moment correlation was conducted at a statistical significance of $p < 0.05$. Statistical analysis was performed using the Statistica 10 (StatSoft, Inc.). Human body angular position was assessed on the basis of chest and pelvis rotation in relation to the coordinate system of the laboratory. All results are given in Table 1.

- Pelvis

The initial value of the pelvis rotation angle in relation to the laboratory coordinate system is influenced by the initial position of the lower limbs. An influence of flexion of limbs in the hips just before releasing the platform lock on the pelvis rotation was observed at the finishing moment of the test recording. The more bended the limbs in both the hip joints, the lower the pelvis extension (turn in the sagittal plane) at the finishing moment of the test recording. Moreover, on the basis of the initial pelvis position, it could be observed that pelvis lateral bending (turn in the frontal plane) to the left (or right) just before releasing the platform lock causes deflection and rotation of the pelvis to the left (or right) in the final phase of a test measurement. A similar situation takes place in the case of rotation (turn in the transverse plane) to the left (or right).

Table 1. Pearson correlation matrix

$N = 28$		Pelvis angle at the finishing moment of a test recording			Chest angle at the finishing moment of a test recording		
		flexion – extension	lateral bending	rotation	flexion – extension	lateral bending	rotation
Initial value of the hip angle	left	$r = -0.46$ $p = 0.013$					
	right	$r = -0.55$ $p = 0.002$					
Initial value of the pelvis angle	lateral bending		$r = 0.51$ $p = 0.006$	$r = -0.46$ $p = 0.015$			
	rotation			$r = -0.55$ $p = 0.002$			
Initial value of the chest angle	lateral bending			$r = 0.37$ $p = 0.05$			
	rotation			$r = -0.54$ $p = 0.003$			
Maximum rotational speed of the lumbar spine	extension				$r = 0.39$ $p = 0.042$		
	rotation to the right					$r = -0.54$ $p = 0.003$	$r = 0.45$ $p = 0.017$
	rotation to the left					$r = -0.58$ $p = 0.001$	$r = 0.45$ $p = 0.017$

where: r – Pearson product-moment correlation coefficient; p – statistical significance; N – number of subjects.

- Chest

The chest rotation angle in relation to the laboratory coordinate system, in the final phase of a test measurement, is influenced by the rotational speed of the lumbar spine. The higher the maximum rotational speed of extension in the lumbar spine section, the higher the angle of the chest extension in relation to the laboratory coordinate system. Moreover, the higher the maximum rotational speed of rotation and lateral bending in the lumbar spine section to the right (or left), the higher the angle of chest rotation and lateral bending to the right (or left) at the finishing moment of a test recording.

4. Discussion

Studies related to human balance are mainly conducted for the rehabilitation purposes. For example, balance rehabilitation can be carried out after stroke [6], lower limbs injuries [8], [25] or for people with different types of disabilities [1], [19]. But there are very few studies in the field of the occupational safety and health described in literature.

The aim of the study was to determine initial conditions describing the movement of the human body at the time of loss of balance. The initial conditions will be used with the numerical human model during reconstructions of a fall from a height. There are no reports in the literature about similar studies related to biomechanics of work. However, numerical human models that use initial conditions (active human model) for the purpose of analysis of car accidents are known. One of the most advanced is active human model developed by TASS-International [15]. The model allows the simulation of human responses during car rollover. The model uses "virtual actuators" to simulate movement of the spine.To develop the model it was necessary to carry out experimental studies with volunteers [18]. The experimental set-up was designed for the quantification of the overall human kinematics (body response) in low-severity impact scenarios. Six male volunteers participated in the tests. In a test set-up a pendulum hits a volunteer laterally in order to investigate the effects of the muscle state on the overall human kinematics. Experiments were captured with the 3D motion analysis system. Moreover, active models of individual parts of the body are developed, an example can be active model of upper extremity using numerical muscles [2], [21]. Another example is an active model of head and neck controlled by a number of muscles used to car accident analyses developed by the University of Loughborough [26].

The idea of conducting a study of the loss of balance process before falling from a height using VR technology was taken from the literature. Effectiveness of this solution has also been proven by numerous researches to treat the fear of heights [3]–[5], [12], [20] and researches for the needs of neurology [11], [13], [24]. The actual test of subject's sensation of being at a certain height has been proven by results of studies by Canadian researchers [5]. They performed an experiment involving comparison of sensations of persons placed at a certain height above the ground with sensations experienced by persons wearing an HMD displaying an image from the perspective of several metres above the ground during a repeated test. The results obtained from the study have proven that the level of fear and the sense of spatial presence during the simulation of being at a certain height above the ground with the use of the VR technology are indeed lower than when actually being at a height, but they increase considerably as compared with standing on the ground without an HMD on.

The conducted tests resulted in obtaining 28 sets of functions describing human reaction at the time of loss of balance. In order to investigate the angular position of the human body, the angle of the pelvis and chest deflection in relation to the laboratory was studied. At the last phase of the recording, all the subjects had their pelvis tilted to the back (at various angles) and showed a tendency to further tilting. What is more, 8 subjects showed deflection and rotation of their pelvis and chest to the right and 7 – to the left.

Movement in the lumbar spine section has a considerable influence on the angular value of backwards deflection. Depending on how strong the flexion or strengthening and its angular velocity are, deflection of the entire body becomes larger or smaller. The degree of the deflection is also influenced by the initial value of flexion of the lower limbs in the hips. It could be inferred that the more a subject was initially (just before releasing the platform lock) leaning forward, the lower was his backwards deflection at the last phase of a test recording and the lower was the deflection velocity. Movement of the upper limbs was not observed to have a significant influence on the trajectory of the recorded part of a fall. This could result from the fact that flexion and abduction of the upper limbs was sudden, but changes of joint angles were small. However, the initial position of a subject on the platform had the greatest influence on deflection and rotation of the body at the recorded moment of a fall. Despite the fact that the subjects' positions were similar, small differences of their positions influenced the course of the fall, frequently causing rotation and deflection of the body. According to the observations, body weight and height did not have an influence on the subject's reaction.

A limitation of the study was the fact that the test recording lasted only 40–65 ms. However, conducting the test on a full scope of a fall from a larger height would be too dangerous for the human subjects. Short time of test recording results also from the use of protection in the form of a harness and a belay rope. A subject falling from a height of 35 cm, at a certain moment, caused stretching of the belay rope, which started to hamper and rotate the subject, distorting the fall trajectory.

As a result of the study, a set of initial conditions was obtained, enabling conducting numerical simulations taking into account human reaction with the use of a human body model from the MADYMO package [23]. This package is used for biomechanical analyses and reconstructions of accidents. As a result, an assessment of human reactions to results of a fall will be possible.

Acknowledgement

Publication prepared on the basis of results of a research task entitled "Study of human reaction at the time of loss of balance with the use of virtual reality technologies" (task III-43), executed within the framework of statutory activity of the Central Institute for Labour Protection – National Research Institute.

References

[1] Błażkiewicz M., *Muscle force distribution during forward and backward locomotion*, Acta of Bioengineering and Biomechanics, 2013, Vol. 15, No. 3, 3–9.

[2] Budziszewski P., Van Nunen E., Mordaka J.K., Kędzior K., *Active Controlled Muscles in Numerical Model of Human Arm for Movement in Two Degrees of Freedom*, IRCOBI Conference Proceedings, International Conference on the Biomechanics of Impact, Bern, Switzerland, 2008, 17–19.

[3] Bush J., *Viability of virtual reality exposure therapy as a treatment alternative*, Computers in Human Behavior, 2008, 24, 1032–1040.

[4] Carmen Juan M., Perez D., *Using augmented and virtual reality for the development of acrophobic scenarios. Comparison of the levels of presence and anxiety*, Computers & Graphics, 2010, 34, 756–766.

[5] Cleworth T.W., Horslen B.C., Carpenter M.G., *Influence of real and virtual heights on standing balance*, Gait and Posture, 2012, DOI: 10.1016/j.gaitpost.2012.02.010.

[6] Cordun M., Marinescu G.A., *Functional Rehabilitation Strategies for the Improvement of Balance in Patients with Hemiplegia after an Ischemic Stroke*, Procedia – Social and Behavioral Sciences, 2014, Vol. 117, 575–580.

[7] Dąbrowski A., *Prace na wysokości – najczęstsze przyczyny wypadków (Works atheights – most frequentcauses of accidents)*, Bezpieczeństwo Pracy Nauka i Praktyka, 2004, 1, 2–6.

[8] Dudek K., Drużbicki M., Przysada G., Śpiewak D., *Assessment of standing balance in patients after ankle fracture*, Acta of Bioengineering and Biomechanics, 2014, Vol. 16, No. 4, 59–65.

[9] Główny Urząd Statystyczny (GUS), Wypadki przy pracy w 2013 r. (Accidents at work in 2013), 2013, GUS, Warszawa, Poland.

[10] GOPRO, Retrieved October 28, 2014, from: http://gopro.com/

[11] Horlings C.G.C., Carpenter M.G., Küng U.M., Honegger F., Wiederhold B., Allum J.H.J., *Influence of virtual reality on postural stability during movements of quiet stance*, Neuroscience Letters, 2009, 451, 227–231.

[12] Krijn M., Emmelkamp P.M.G., Biemond R., De Wilde De Ligny C., Schuemie M.J., Van Der Mast C.A.P.G., *Treatment of acrophobia in virtual reality: The role of immersion and presence*, Behaviour Research and Therapy, 2004, 42, 229–239.

[13] Lee H.-Y., Cherng R.-J., Lin CH.-H., *Development of a virtual reality environment for somatosensory and perceptual stimulation in the balance assessment of children*, Computers in Biology and Medicine, 2004, 34, 719–733.

[14] MADYMO Software Documentation, *MADYMO Human Body Models Manual Release 7.5*, 2013, Tass-International.

[15] Meijer R., Rodarius C., Adamec J., Van Nunen E., Van Rooij L., *A first step in computer modelling of the active human response in a far-side impact*, International Journal of Crashworthiness, 2008, 13(6), 643–652.

[16] Milanowicz M., Budziszewski P., *Wykorzystanie komputerowego modelu człowieka do rekonstrukcji wypadków przy pracy (Application of a computer model of the human body for reconstruction of accidents at work)*, Mechanik, 2011, 7, 567–574.

[17] Milanowicz M., Budziszewski P., *Numerical Reconstruction of the Real-Life Fatal Accident at Work: A Case Study*, Lecture Notes in Computer Science (V.G. Duffy (ed.): DHM/HCII 2013, Part II), 2013, 8026, 101–110.

[18] Muggenthaler H., Von Merten K., Peldschus S., Holley S., Adamec J., Praxl N., Graw M., *Experimental tests for the validation of active numerical human models*, Forensic Science International, 2008, 177, 184–191.

[19] Nardone A., Godi M., Artuso A., Schieppati M., *Balance Rehabilitation by Moving Platform and Exercises in Patients With Neuropathy or Vestibular Deficit*, Archives of Physical Medicine and Rehabilitation, 2010, Vol. 91, Issue 12, 1869–1877.

[20] Olasov Rothbaum B., Hodges L., Alarcon R., Ready D., Shahar F., Graap K., Pair J., Hebert P., Gotz D., Wills B., Baltzell D., *Virtual Reality Graded Exposure in the Treatment of Acrophobia: A Case Report*, Behavior Therapy, 1995, 26, 547–554.

[21] Östh J., Brolin K., Happee R., *Closed loop control of FE arm model*, IV European Conference on Computational Mechanics Palais des Congrès, Paris, France, May 16–21, 2010.

[22] Razer, Retrieved October 28, 2014, from: http://www.razerzone.com/

[23] Tass International, Retrieved October 28, 2014, from: http://www.tassinternational.com/madymo

[24] Tossavainen T., Juhola M., Pyykkö I., Aalto H., Toppila E., *Development of virtual reality stimuli for force*

platform posturography, International Journal of Medical Informatics, 2003, 70, 277–283.

[25] TRUSZCZYŃSKA A., RĄPAŁA K., GMITRZYKOWSKA E., TRZASKOMA Z., DRZAŁ-GRABIEC J., *Postural stability disorders in patients with osteoarthritis of the hip*, Acta of Bioengineering and Biomechanics, 2014, Vol. 16, No. 1, 45–50.

[26] VAN LOPIK D.W., ACAR M., *A computational model of the human head and neck system for the analysis of whiplash motion*, International Journal of Crashworthiness, 2004, 9(5), 465–473.

[27] Vicon, Retrieved October 28, 2014, from: http://www.vicon.com/