Kinematic Analysis of Backward Falls of Pedestrian and Figurine in Relation to Head Injury

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ABSTRACT: Unexpected backward falls caused by a vehicle at a pedestrian crossing or at a supermarket parking space, by slipping on the road or by an external force acting against the chest followed by a backward fall is a very frequent phenomenon and can affect anyone. Especially the back and temporal part of the head come into contact with the ground during this uncontrolled motion and, depending on the impact strength and surface properties, it can cause serious injury. This study addresses a kinematic analysis of probands' backward fall (voluntary fall, chest impact fall, slipping fall) and correlates it with dummy-based data and subsequently with data from head-surface interaction. From the measured head acceleration we can therefore calculate head injury criteria related to individual kinematic analyses and values during the fall.

KEY WORDS: Backward fall, head injury, kinematic analysis, pedestrian, Qualisys.

1 INTRODUCTION

Falls can be grouped as standing falls, height falls and free falls. This study addresses standing falls, which are the most common, affecting a wide proportion of population. The standing falls is a threat especially to children, elderly people and alcohol-affected people. According to Deemer et al. (2005), falls are a major cause of morbidity and mortality in children, but are also reported falsely in child abuse. Therefore, it is of interest to understand those factors which may lead to a higher likelihood of injury in a feet-first freefall. Seniors are hospitalized twice as often as the general population for fall-related traumatic brain injury (TBI), while over half of all fall-related deaths in older adults are due to TBI. The risk for fall-related TBI increases substantially with age; persons over the age of 85 are hospitalized for fall-related TBI over twice as often as those aged 75–84, and over 6 times as often as those aged 65–74 (Manavais et al., 1991). In the conclusion of their findings, Kool et al. (2008) suggest that drinking in the previous 6 h has a strong and consistent relationship with the risk of unintentional falls.

Standing falls can be further grouped by direction into backward falls, forward falls, sideward falls and combined falls. Injuries due to backward falls, apart from sideways fall, are a major health problem, particularly among the aged populations (Majumder et al., 2009). A total of 327 injuries were identified in 259 patients. Back injury was the most common (19.3%) (Smith & Nelson, 1998).

The next category of falls are falls after car impact. The impact speed of the striking car is widely accepted as a prime factor for the injury risk in car-to-pedestrian collisions. However, people's opinions differ as to the exact relationship between car impact speed and pedestrian fatality risk (Rosén et al., 2011). According to Rosén and Sander (2009) a strong dependence on impact speed is present, with the risk at 50 km/h being more than twice as high as the risk at 40 km/h and more than five times higher than the risk at 30 km/h. A lot of studies are concentrated on the reconstruction and modeling of pedestrian crashes at speeds higher than 20 km/h, but 12.1% of pedestrian injuries are caused by car speeds less than 10 km/h (Peng et al., 2012).

Slip and fall accidents have been recognized as a major threat to the safety of individuals, not only in industry, but also in daily life. According to the 2002 annual report of science activity, “same level fall” and “fall to lower level” were cited as two of the five leading injury causes accounting for 5 or more days away from work (Yoon & Lockhart, 2006). Mechanical impact is the leading cause of injury, death and disability in people aged under 45 in the USA, Europe, and, increasingly so, in Third World countries (Jennett, 1996). In Ireland, falls are the single greatest cause of hospital admissions for both males and females across most age groups, with head injuries occurring in approximately a quarter of fall admissions.

After contact between the head and a surface multiple injuries take place. From an injury mechanics point of view both direct injury, caused by the impact of the head on a surface, and indirect injury are concerned. The brain can therefore be injured locally - acceleration-based injury (so called translation injury). The primary injury damage of the brain usually corresponds to a point of impact, but can also take place contra-laterally by means of par contre-coup mechanism. Apart from lacerations of head skin the bony part is injured too - neurocranium fractures can be divided into fissures, comminuted fractures and traumatic brain injuries (TBI). Traumatic brain injury (TBI) is an important cause of injury-related hospitalization, disability, and death worldwide. It is of particular concern in the older population as functional recovery following an acute injury to the brain is often limited and can signal the end of independent living (Harvey & Close, 2012). The tolerance of the head to skull fractures is much easier to determine than its tolerance to intracranial injury. This is because of the definite relationship between force applied to the skull, and failure of cranial bone (Wright & Laing, 2012). In cases of severe head injury caused by a fall, coup contusions are either absent or very minor, in contrast to the presence of extensive contre-coup damage. In cases of a severe blow to the head, however, the reverse occurs, with contre-coup lesions a rarity and coup damage extensive (Yanagida et al., 1989). A study examining the pattern of brain injuries in falls resulting in death found skull fractures in 66% of cases. Acute subdural haematoma (ASDH) have been found in between 78% and 85% of fatal falls (Manavais et al., 1991).

There are three most common approaches to biomechanical analyses. Experiments involving human participants, dummies and computer simulations. Analyses of falls with human probands were carried out by Klenk et al. (2011) using two measurement protocols:

- Fall simulation was conducted onto protective layers of mattresses. Participants stood at a distance of 1.5 times the lengths of their foot apart from the mattresses with their back to the mattress and were instructed to “fall backwards as if you were a frail old person”.
- The participants now were instructed not to fall onto the mattresses, if possible, when released from a backward lean. The instruction was “when we release you, try as hard as you can not to fall”. The participants were held by a staff member in a backward lean of about 30–40°. The inclination of the body position was adjusted so a fall was unavoidable.

Computational simulation of real life head injury accidents has been used for various purposes. Some have compared AIS (abbreviated injury scale) scores for real life injuries to HIC scores or other indices of injury calculated from the reconstruction (O'Riordain et al., 2003). Evaluation of head impact dynamics is commonly accomplished using mechanical impact simulators. Such tests have found widespread use in the development of safety standards for devices including helmets, airbags, and playground surfaces (Wright & Laing, 2012).

An optimal approach turns out to be a combination of human probands and dummies. The conditions are set among the probands and measurements are done within the limits of human tolerance. Dummy measurements follow along with a validation of input data and a simulation of extreme values. This data can then be implemented in computer models.

2 MATERIAL AND METHODS

Klenk et al. (2011) used only the accelerometer data that had been affected by mattress impacts. In our study we used a Qualisys system for both probands and dummies for a comprehensive kinematic analysis of the fall onto a mattress, but for assessment of possible head injuries the Manikin dummy, equipped with a three-axis accelerometer in the head, was made to fall on a flat, dry surface of asphalt concrete (ACO 11). We speculate two possibilities for the cause of the fall. The first one, that cars on the pedestrian crossing are not able to stop and hit the pedestrian with a small velocity (up to 10 km/h) or a car pulls away just at the moment when pedestrian enters its path and hit him by a small velocity (up to 10 km/h) (Fig. 1). In the second case, the person falls after an impact to the chest (Fig. 2).

Figure 1: Backward fall of dummy after car impact.

Figure 2: Backward fall of dummy after chest impact.

As the methodics of probands falls we analyzed three ways of performance. Voluntary backward fall with an instruction to the proband to fall as erect as possible, an impact to the chest without the possibility to step backwards and finally a fall after pulling the legs forward (Fig 3).

Figure 3: Tree types of backward falls.

To assess the severity of head injury a maximum acceleration, 3-ms criterion a HIC_{36} were chosen as comparison criteria. 3-ms criterion is applicable not only to head injury. The limiting value is 80g and it means that an acceleration higher than 80g must not last over 3 ms. The formula for the HIC is defined as:

$$HIC = \left\{ (t_2 - t_1) \left(\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) dt \right)^{2,5} \right\}_{max}$$

where t_1 and t_2 are, respectively, the start and finish times of the acceleration spike, and $a(t)$ is the resultant linear head acceleration with respect to time. HIC values were calculated over 36 ms (HIC_{36}) durations to allow comparisons of results with existing injury criteria. Head injury criterion HIC use is based on a proposal of the National Highway Traffic Safety Administration (NHTSA), 1972 (Marjoux et al., 2008). For the effects of direct impact

it has been demonstrated that HIC is an acceptable discriminator between severe and less severe injuries (Tarriere, 1981). It also correlates with the risk of fractures to the skull. According to EHK 94 regulation (First, 2008) the threshold is then determined as HIC = 1000 and acceleration that is greater than 80 g for no longer than 3 ms. HIC $_{36}$ was designed to protect the head against injuries, such as fractures of the skull with a longer exposure when there is no contact with the hard parts of the interior.
The dummy Manikin and three healthy men in their middle ages (age 29 ± 3 and 75 ± 7 kg) were tested. Multi-analyzer system Dewetron, 3 measuring channels and one output channel and accelerometer MMF KS 943 with a sampling frequency 5000 Hz were used for the measurement of acceleration. The motion capture system Qualisys with 6 cameras and 400 Hz sampling frequency was used for scenery capturing and Qualisys Track Manager was used for the motion capture process. The software package HyperWorks (application HyperView and HyperGraph) was used for evaluating the measured data. The data were filtered after import according to the EuroNCAP methodology (EuroNCAP, 2011).

3 RESULTS

The kinematic data were evaluated using a Qualysis Track Manager. Overall results can be seen in Table 1. The following parameters were analyzed: maximum speed (impact speed) on the top of the head - in Os Parietale area, on the front of the head on Os Frontale, on both shoulders, furthermore the travelling distance of the Os Parietale spot and, thereby, the overall change of head rotation against the shoulders. There were 8 measurements of the figure – three were after chest impact, five were after car impact. There were no statistical differences between chest and car impacts.

Table 1: Kinematic data.

		Max. velocity Head_ Parietale (mm/s)	Max. velocity Head_ Frontale (mm/s)	Max. velocity Shoulder_R (mm/s)	Max. velocity Shoulder_L (mm/s)	Distance Head Parietale (mm)	Difference of angle (^{o})
Dummy	Mean	6018,6	5078,0	4755,0	5328,8	3003,0	5,1
	SD	140,0	175,5	99,8	91,3	117,9	0,1
Backward Fall	Mean	5 126,4	4 542,8	4 102,7	4 953,9	2 973,8	5,4
	SD	511,8	376,9	185,2	796,9	79,5	0,1
Backward Fall - chest force	Mean	5 716,5	4 965,3	4 812,9	5 071,9	3 072,8	5,7
	SD	290,3	202,6	512,7	630,1	48,7	0,3
Backward Fall - leg slide	Mean	5 289,1	4 596,9	4 748,2	4 753,8	2 708,4	18,7
	SD	481,5	336,8	551,5	484,2	63,2	1,8
T-Test Dummy vs. Backward Fall		0,10	0,07	0,08	0,61	0,85	0,05
T-Test Dummy vs. Backward Fall - chest force		0,17	0,10	0,88	0,57	0,61	0,05

The data from the accelerometers were imported into the software HyperGraph and filtered according to the method of Euro NCAP – filter CFC 1000. CFC 1000 is characterized by the following parameters: 3 dB limit frequency is 1650 Hz, stop damping is -40 dB, and sampling frequency is at least 10 kHz. The values of recorded accelerations in three axes were converted to the resultant acceleration of the head and the values of the HIC, 3ms criterion, and the maximum acceleration were computed. Results are shown in Table 2.

Table 2: Data from accelerometers.

	Max. a (g)	3-ms (g)	HIC 36	Max.velocity Head_Parietale (mm/s)	Max.velocity Head_Frontale (mm/s)
Dummy 1	120	45	794	6 427,1	6 130,0
Dummy 2	265	47	816	6 608,8	6 328,2
Dummy 3	225	56	911	6 734,2	6 259,7
Dummy 4	314	52	895	6 724,8	6 430,5
Dummy 5	198	51	843	6 879,0	6 425,0
Dummy 6	212	52	899	6 898,0	6 489,0
Dummy 7	112	41	698	6 245,0	5 890,0
Dummy 8	324	61	935	6 987,0	6 423,0

4 DISCUSSION

The gathered data were imported into the STATISTICA CZ 10 software, where they were statistically evaluated.

It can be stated that the dummy and the probands can be compared on a significance level 0.05. It is therefore possible to consider the dummy's behavior during the backward fall comparable to a human body. The closest to a dummy are the values of a fall after chest impact (see Fig. 4). This is because the dummy does not bend during the fall, while the conscious human reflexively bends the torso and dampens the fall, resp. decreases the impact speed.

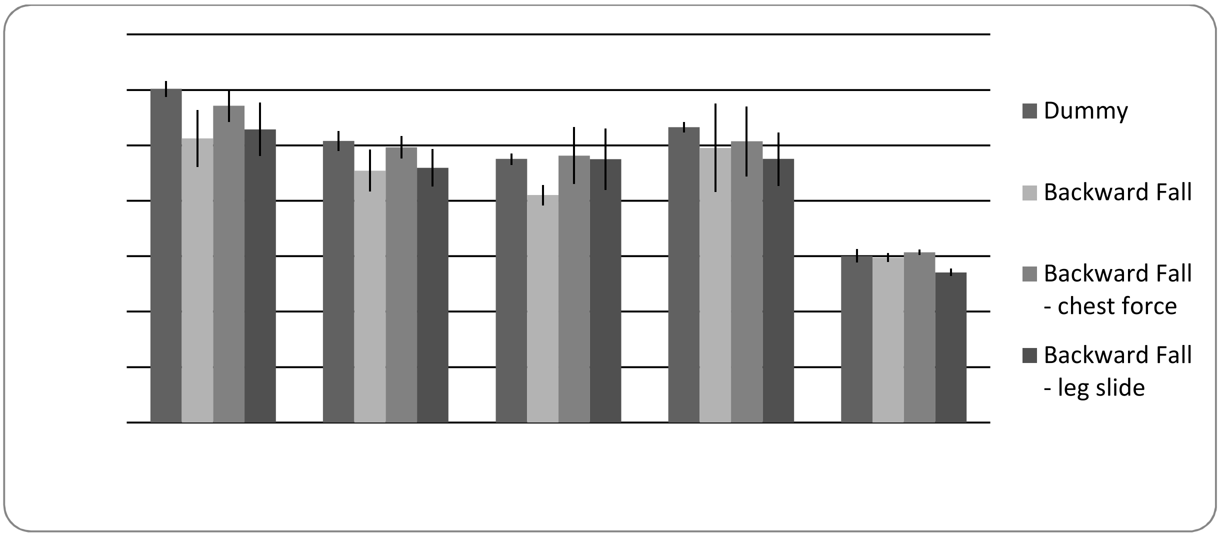

Figure 4: Kinematic data in Graph.

During the voluntary backward fall there was a strong correlation found between top speeds of Os Frontale and Os Parietale (r=.998), which is understandable. What is interesting is no proven correlation between the rest of the body segments. Only the distance travelled was close to the top speed on the Os Parietale (r=.0,942).

If we look at the correlations of individual head injury criteria with impact speeds (Table 4), we can find interesting data for projecting a possible head injury from the fall type and the impact speed (Fig. 5).

Table 3: Injury correlations.

	Max. a	3-ms	HIC 36	Max.velocity Head_Parietale	Max.velocity Head_Frontale
Max. a	1,00	0,86	0,85	0,77	0,72
3-ms	0,86	1,00	0,94	0,88	0,72
HIC 36	0,85	0,94	1,00	0,90	0,85
Max.velocity Head_Parietale	0,77	0,88	0,90	1,00	0,92
Max.velocity Head_Frontale	0,72	0,72	0,85	0,92	1,00

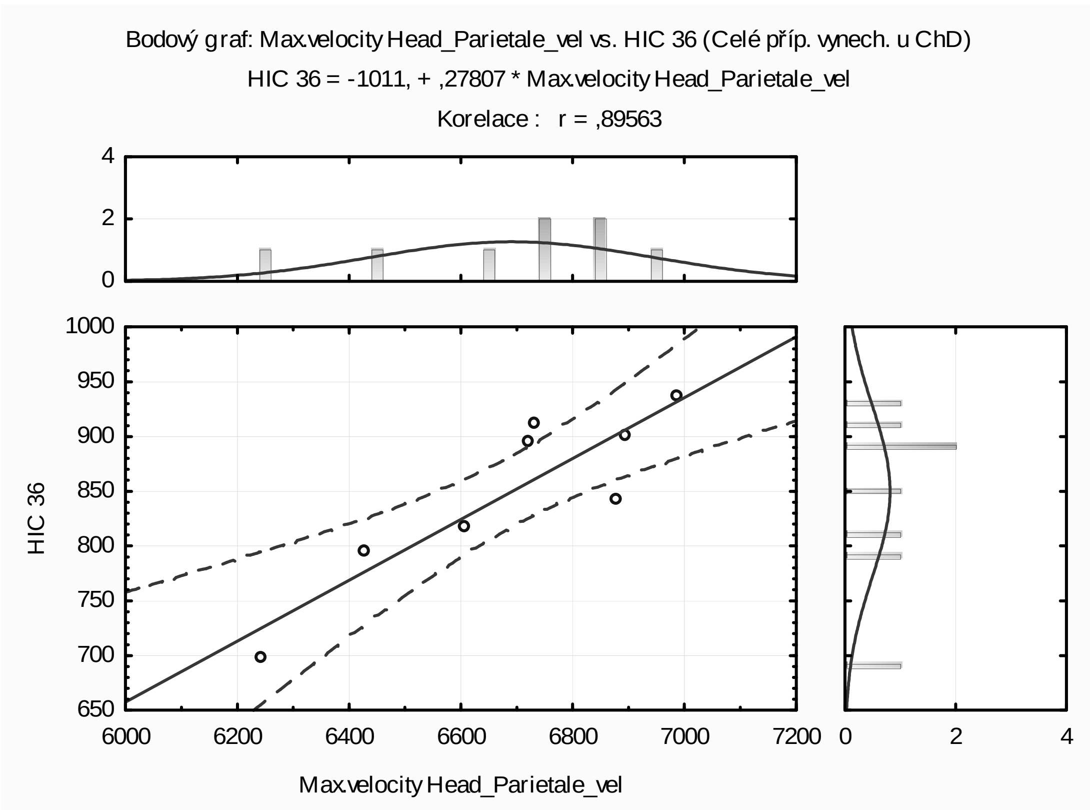

Figure 5: Max. velocity and HIC 36 correlation.

Impact surface significantly affected injuries. Since harder surfaces deform less during an impact they result in a more rapid deceleration when compared with a softer surface. Consequently, the fall energy must be absorbed over a shorter period of time for harder surfaces than softer surfaces. This allows the body less time to distribute and dissipate the energy, potentially resulting in a higher injury risk. The friction of the impact surface may also affect injury risk. Surface friction is typically reported using the coefficient of friction (COF). The higher the COF, the greater the surface friction. Whenever shear forces exceed the COF of the shoe-floor interface, a slip can result (Deemer et al., 2005).

Results of studies about pedestrian fatality risk as a function of car impact speed (Anderson et al., 1997; Tefft, 2012) say that there is a very small probability of severe injuries or death. But reality could be different. When a car hits a pedestrian at a minimal speed and the pedestrian does not expect the impact, he then falls with uncontrolled motion and, depending on the impact strength and surface properties, this can cause serious injury. This is the reason why we must analyze isolated falls as well.

If we look at a reconstruction of real world head injury accidents resulting from falls using multibody dynamic (O'Riordain et al., 2003), there is an input speed unknown variable which can change results very dramatically. Klenk et al. (2011) presented the acceleration pattern of a real-world fall and simulated fall. But acceleration is only from the centre of gravity (of the whole body) and only during the fall phase, not during impact. The influence of headform orientation and flooring systems on impact dynamics during simulated fall-related head impacts was measured by Wright and Laing (2012). He used a mechanical head impact simulator at 1.5, 2.5, and 3.5 m/s impact velocities, which are too low according to our study. Our results, especially for head impact velocity, could be very useful for simulations in the future.

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