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RECONSTRUCTING REAL LIFE ACCIDENTS TOWARDS ESTABLISHING CRITERIA FOR TRAUMATIC HEAD IMPACT INJURIES

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Abstract. Brain injury is the leading cause of death in those aged under 45 years in both Europe and the United States. The objective of this research is to reconstruct and analyse real world cases of accidental head injury, thereby providing accurate data which can be used subsequently to develop clinical tolerance levels associated with particular traumatic injuries and brain lesions. In this study, MADYMO pedestrian models are used to analyse a well-defined set of non-fatal accidents involving simple falls. The effect of varying the initial conditions is systematically examined and the predicted MADYMO results are compared against literature data.

Key Words: Impact biomechanics, falls, accident reconstruction, head injury, multibody dynamics.

1. INTRODUCTION

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 [1]. Costs of hospitalisation, care and rehabilitation of head injured people are estimated to be as high as \$33 billion per year in the USA [2]. Much recent research has focussed on the biomechanics of traumatic head injury, an objective of which is to correlate clinical dysfunction with mechanical impact conditions, with a view to reducing or eliminating the mechanisms that cause damage.

Road traffic accidents (RTAs), falls and assaults are the most frequently cited causes of head injury. While RTAs tend to be the leading cause of injury related death, falls tend to be the leading cause of non-fatal hospitalisation [3,4]. 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 all admissions [4]. Falls are selected as the accidents of interest in this present study due to their lower levels of uncertainty regarding initial conditions, and due to the tendency of falls to result in focal, as opposed to diffuse, head trauma.

The aim of this research is to reconstruct a number of real life falling accidents using numerical techniques in order to provide detailed data which can be used to establish injury criteria for specific types of brain lesion. The accidents are modeled using multibody dynamics software to recreate the overall movement of the body during the accident. Once this simulation represents the accident described, the output, in the form of velocities, accelerations and forces, is subsequently used as input for a

3D finite element model of the head, which has been previously compared with experimental results [5]. This finite element model is able to simulate the effect of the overall head movement on the cranial contents, so the local deformation parameters within the brain tissue can be examined and compared to the observed clinical results. This present paper only presents the results of the multibody dynamics simulations and compares the results against those reported by other researchers.

2. METHODOLOGY

Real life cases of falls resulting in head injury are selected by Ireland's National Department of Neurosurgery at Beaumont Hospital, Dublin. Cases are screened to narrow the selection to relatively simple falls, in order to facilitate modelling of the accidents. Clinical assessments of each case are provided by the hospital, together with CT scans. The accident site is examined to determine the layout of the environment, the height of the fall, and the type of surface onto which the person fell. Informed consent is obtained from patients and witnesses, with the approval of the Ethics Committee of Beaumont Hospital.

Accident reconstruction is carried out using MADYMO (MAThematical DYnamic MOdels) [6] multibody dynamics software. MADYMO has a database of dummy models, which makes it very suitable for reconstructing accidents involving humans. For these analyses the pedestrian models, which have been validated extensively against full body pedestrian tests, were used, with altered head contact characteristics. It was found that the values for the forces and accelerations experienced by the head of the pedestrian model were very high in comparison to values cited in the literature. From previous related research [7] it was determined that the head response curve determined by Yoganandan *et al* [8] was the most suitable for this analysis since it was independent of the head impact location.

In order to reconstruct these accidents certain initial conditions need to be applied to the model. Due to the fact that all data regarding the accidents were collected from the field, there is necessarily a degree of uncertainty regarding the precise conditions under which each accident took place. Without instrumentation attached to the person involved, it is impossible to know exactly the velocities of the people during the fall. The cases presented here have at best an eyewitness account of the accident, which is of course useful and necessary, but not scientifically rigorous. Initial conditions in MADYMO are defined by specifying the X, Y and Z components of both linear and angular velocity, and initial joint rotations and positions. For each case reconstructed here, an estimate was made of these components of the initial velocities and positions, based on available information regarding the accident. These were used to run an initial test simulation. If the graphical representation of the simulation appeared unrealistic, slight alterations were made to the initial velocities until the kinematics of the impact appeared physically

realistic and correct. This method still leaves some uncertainty as to the validity of these initial conditions. In order to systematically consider the effect of these values on the results, the initial conditions were then varied by $\pm 10\%$ and $\pm 50\%$. Figure 1 shows a sequence of events taken from one of the accident simulations. The relevant results obtained from these simulations are in the form of linear and angular accelerations (a , \dot{g}), velocities (v , \dot{g}) and forces experienced by the head.

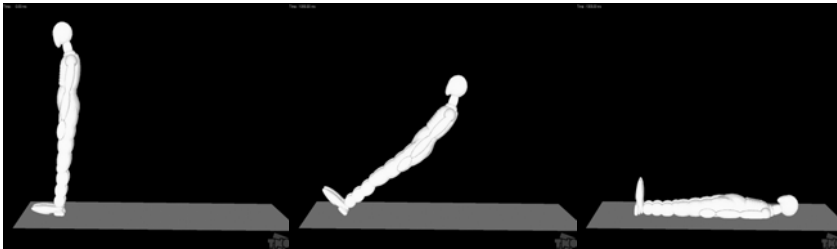


Figure 1. Images taken from the simulation of a boy falling at a water fountain

3. RESULTS

In this study a total of ten real-life accident cases were analysed. A summary of these cases is given in Table 1. For each of the cases a sensitivity analysis was performed. This was done by varying both the initial joint positions and the initial joint velocities, both on their own and combined, by $\pm 10\%$ and $\pm 50\%$. Some of the simulations resulting from these changes in initial conditions did not accurately represent the kinematics of the accident described, and so these results were omitted from the subsequent analysis. In some cases where there was no initial joint velocity, a certain amount of velocity was applied to the body in the direction of the movement of the fall. For example, in Cases 1 and 2, the people fell directly backwards from stationary positions. These cases were modelled by tilting the body backwards and allowing it to move under the influence of gravity. The sensitivity analysis looked at the effect of applying a small amount of backward rotational velocity to the body to see how this affected the results. In some of the cases there was very good agreement in the results despite changing the input by up to 50%. This was particularly evident in the cases where the fall was in one plane only, i.e., the person fell straight forwards or straight backwards. In the cases where there was out-of-plane motion, greater differences could be seen in the results. However, if the initial conditions were changed by too much for these accidents it was more likely that they would no longer represent the kinematics of the real-life accident accurately. Table 2 shows the range of results obtained from the simulations representing the accidents. In general there is good agreement among the results. The case with the largest differences is Case 9. This involved a man falling from a gate at a height of 138cm above standing height, giving more time for voluntary reactions by the person. The simulations do not take any voluntary reactions into account, therefore there is likely to be a

higher degree of error in these results, and this is evident by the wider range of results observed for the same accident description.

Table1. Case summaries, providing description of accident and head injury sustained.

Case	Sex	Age yrs	Height cm	Weight kg	Brief description of accident	Head injury
Case 1	F	76	160	60	Lost her balance and fell directly backwards. Incurred occipital impact of head against concrete wall.	Small left frontal lobe contusion. Large right temporal parenchymal haematoma.
Case 2	M	11	152	37	Fell directly backwards. Incurred occipital impact of head on concrete pavement.	Right frontal lobe contusion. Traumatic subarachnoid haemorrhage.
Case 3	M	85	163	70	Fell directly forwards after losing his balance onto concrete footpath.	Left sided chronic subdural haematoma. Right sided acute subdural haematoma. Midline shift to the left.
Case 4	F	84	163	63.5	Fell directly forwards after losing her balance while walking downhill on concrete footpath.	Left sided subdural haematoma with midline shift. Dilated right ventricle.
Case 5	F	84	163	57	Tripped on a crack causing her to fall forward and to her right, hitting head off concrete footpath.	Right sided acute and chronic subdural haemorrhage with midline shift and subfalcine herniation.
Case 6	F	71	163	63.5	Lost her balance and fell forwards hitting the front left of her head off the concrete ground.	Left fronto-parietal subdural haematoma with midline shift and asymmetrical ventricles.
Case 7	M	76	170	66.7	Tripped causing him to fall forward on his right side, breaking his shoulder and hitting right side of his face.	Right chronic subdural haematoma. Left acute subdural haematoma.
Case 8	F	87	157	51	Slipped on a ramp and fell forward and hitting the right of her head off a railing.	Left sided subdural haematoma. Lateral ventricle shifted to the right side.
Case 9	M	24	169	80	Fell backwards and twisted to the left while balancing on a gate, pulling a rope which broke. Incurred left lateral impact of head on tarmac.	Left temporo-parietal linear skull fracture. Left temporo-parietal extradural haematoma.
Case 10	F	24	169	55	Standing on chair, twisted sharply and fell forwards and to the right. Incurred right lateral impact of head on ceramic tiled floor.	Right frontal linear skull fracture. Right frontal extradural haematoma. Left posterior temporal contusion.

The cases where the results of the sensitivity analysis are in closest agreement are those for which the accident occurred in one plane only (i.e., Cases 1-4), resulting in either frontal or occipital impact. For two of these simulations Figure 2 shows the envelope of maximum and minimum values for the range of results. It can be seen that the

difference between these curves is modest, with the greatest difference occurring for the accelerations of Case 1 ($\pm 21\%$). In this case the lady's head impacts the wall behind her, and not the ground, so slight differences in head position are likely to lead to larger differences in acceleration values. Figure 3 represents the accidents where the impact was fronto-parietal. There is a lot more variation in these results than for the planar impacts (greatest difference is $\pm 46\%$).

Table2. Summary of the peak results for the sensitivity analyses.

Case	v (m/s)	$\dot{\theta}$ (rad/s)	a (g)	$\ddot{\theta}$ (krad/s ²)	Force (kN)	HIC
1	4.83 - 5.75	32.14 - 38.9	195 - 300	30.0-44.5	6.9-10.38	511.83-1200
2	6.34 - 7.2	44.05 - 49.44	333 - 403	33.9 - 42.0	12.67 - 14.7	2930 - 4308
3	5.08 - 5.64	21.1 - 24.16	321 - 365	13.2 - 16.1	15.43 - 16.88	2491 - 3255
4	4.5 - 4.77	16.22 - 20.38	343 - 364	19.9 - 27.8	11.79 - 13.03	2387 - 2831
5	3.51 - 4.21	13.62 - 19.21	242 - 298	14.9 - 17.9	8.75 - 10.96	1026 - 1786
6	3.61 - 3.67	17.62 - 19.43	302 - 307	23.8 - 24.4	11.19 - 11.31	1269 - 1312
7	4.73 - 5.11	12.12 - 17.56	303 - 330	7.4 - 15.1	9.85 - 11.84	1942 - 2408
8	5.41 - 6.11	58.83 - 72.91	265 - 351	41.9 - 49.2	11.25 - 12.64	1532 - 2914
9	6.9 - 8.19	15.52 - 26.78	189 - 456	8.1 - 22.1	8.8 - 22.11	674 - 5951
10	5.74 - 6.6	21.06 - 35.29	318 - 342	23.7 - 30.2	10.88 - 11.65	2237 - 2639

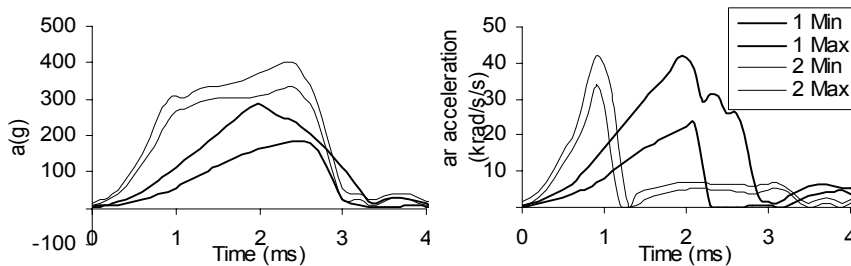


Figure 2. Maximum and minimum values obtained from the range of simulations for the falls that occurred in one plane only. (Cases 1 and 2)

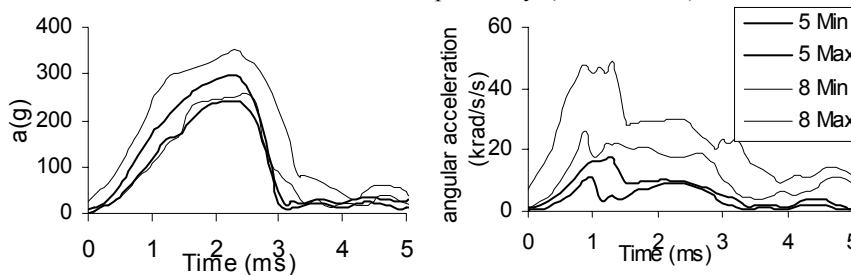


Figure 3. Envelopes of maximum and minimum values predicted for the falls that occurred in more than one plane. (Cases 5 and 8)

Examining the simulations where the fall is out of plane (Cases 5-10), it can be seen that there is greater variability in the results. In these cases it is more likely that by changing the initial conditions significantly the kinematics of the simulation will change to such an extent that it no longer matches the accident description. In some of these cases the

results are omitted since even a slight change in kinematics can lead to quite a large change in the resulting velocities, accelerations and forces.

4. DISCUSSION

In this set of accident cases there were two occipital impacts (Cases 1 and 2). In both of these cases contre-coup contusion was observed on the frontal lobe of the brain, with no evidence of coup contusion present. One other case involving lateral impact also presented with contusion. In all three cases where contusion was observed the linear acceleration falls above the lower tolerance curve for contusion proposed by Auer *et al* [9] indicating that contusion is a possibility. In fact, all cases observed in this study have linear accelerations above this level. However there is no upper tolerance curve provided for contusion, so there is no level given above which contusion will always be observed. In some of the other cases there may be contusion present, but it may be masked by the presence of a subdural haematoma.

In Case 1 parenchymal haemorrhage was also observed. Very little research has been carried out on the biomechanics of this specific type of lesion. One could argue that a minimum level of acceleration is necessary for parenchymal haemorrhage to occur. It is likely that it is associated with higher accelerations than contusion, as it is a more severe injury. This case has very high angular accelerations, due to the oblique nature of the impact, and it is possible that this contributes to the impact of the brain against the sphenoid ridge, which may contribute to the occurrence of both contusion and parenchymal haemorrhage.

Subarachnoid haemorrhage can be seen in Case 2 as bleeding in the right Sylvian fissure. In this case there is high linear acceleration and linear velocity. Bleeding in the Sylvian fissure may have been caused by damage to the blood vessels in the middle fossa region of the brain arising from this very high linear acceleration. It must be noted that this case involves an 11 year old boy. The skull does not become fully calcified until adulthood and Mohan *et al* [10] suggest that at age 13, skull stiffness is only 90% that of an adult. If this factor were taken into account in the force penetration curves for head contact, lower forces should be predicted on the head.

There are also two cases involving lateral impact in this study (Cases 9 and 10). Both of these accidents involve a fall from a height higher than standing height. Both patients were 24 years of age at the time of the accident, and both sustained linear skull fracture and extradural haematoma. Fracture occurred at the location of impact with underlying extradural haematoma. Yoganandan *et al* [8] found that the force necessary to fracture cadaver skulls ranged between 8.8 kN and 14.1 kN, with an average of 11.9 kN. A similar study by Allsop *et al* [11] for temporo-parietal bone found an average of 12.4 kN. Peak forces in these cases were sufficient to cause skull fracture. Extradural haematoma involves the damage of blood vessels lying between the dura and the

skull. Blumbergs [12] cites studies suggesting that skull fracture is associated with extradural haematoma in up to 97% of cases. The skull fracture and extradural haematoma can be attributed directly to contact effects (i.e., force and linear acceleration), which were quite high in these two cases.

The majority of the cases in this study involved frontal or fronto-parietal impacts. In all these cases subdural haematoma was observed, often accompanied by shifting of the midline and/or the ventricles. The location of the subdural haematoma had no apparent correlation with where the impact occurred, but was observed most often on the contrecoup side. No other injuries were observed in these cases, however there may be contusion underlying the SDH in some cases. Auer *et al* [9] produced upper and lower tolerance curves correlating subdural haematoma with linear acceleration. All of the present cases had linear accelerations that were between these lower and upper tolerance curves, which indicates that subdural haematoma is a possibility but not a certainty. Other authors [13] suggest that high angular acceleration together with a high strain rate (i.e., rate of acceleration onset) are the important factors in ASDH causation. They suggest that falls resulting in a peak head acceleration of over 200g and a duration of 3.5ms or less would produce the conditions necessary for the occurrence of ASDH due to bridging vein rupture. These conditions are met in all cases presented here with high accelerations and very short duration impacts (in the region of 3ms). The likelihood of subdural haematoma due to bridging vein rupture is also thought to increase with age, and in all cases presenting with this type of injury the age of the patient is quite high.

Looking at the simulations overall there is no obvious effect of increasing or reducing initial joint velocities or positions. Each case reacts differently. In general, changing the initial velocity seems to have more of an effect on the results than changing initial joint positions and rotations. In many cases changing the initial joint rotations alters the simulations to such an extent that it no longer corresponds to the kinematics of the accident described.

5. CONCLUSION

This analysis has shown that multibody modelling is a useful tool for reconstructing real-life accidents. From the results presented here it can be seen that the accuracy is appreciably greater when simulating simple accidents than more complex accidents. Falls from standing height generally gave good agreement in the results once the kinematics of the simulations represented the accident well. However, in the case where the man fell from the gate there was greater variation in the results. It is quite likely that this difference is partly due to the fact that there is more time and space for the person to react when they are falling, and voluntary reactions cannot be taken into account by the MADYMO software. This will inevitably lead to errors in the results and a wider

envelope of results for a given range of input conditions. The more details that are known about the accident the more accurate the simulations will be and the less scatter will be associated with the results.

The main disadvantage of using real-life accidents is that most of the inputs rely on eyewitness reports which are often not accurate. However, in this study it can be seen that there is a limited range of input conditions that will result in a kinematically realistic simulation of the accident, and the outputs of these simulations generally agree quite well.

The main advantage of using real-life accidents is that the injuries are known. If the initial conditions are accurately reported and the injuries are known, it should be possible to see which kinematic inputs lead to particular types of brain lesions. Ongoing work in this study involves the use of finite element modelling procedures. The results from the multibody modelling, in the form of velocities, accelerations and forces, are being used as input for a 3 dimensional finite element model of the head. This gives output in the form of brain tissue deformation resulting from the head impact. This information will be compared in a subsequent study against clinical data in order to establish quantifiable mechanical thresholds for the occurrence of different types of trauma.

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