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3D multibody dynamics analysis of accidental falls resulting in traumatic brain injury

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Abstract - Clinical, physical and mechanical details of one of a set of ten real world accidental falls which resulted in non-fatal head impact injury in the form of various traumatic brain lesions are presented herein. These are analysed and described in depth in the present paper, along with the accompanying time profiles of linear and angular velocities of all ten accident cases, which were predicted using multibody dynamics modelling simulations. It is suggested that these cases could usefully constitute the basis of a database of documented head injury cases that may be of use to the wider research community. As such, these are freely available to researchers who would wish to use this set of data, upon direct request to the authors.

Keywords: Brain injury; accident; fall; multibody dynamics modelling

INTRODUCTION

The mechanics of neurotrauma are most commonly analysed by means of accident reconstruction techniques, controlled laboratory experiments using either instrumented anthropomorphic human dummies or anaesthetised animals, or computational simulations using multibody dynamics models, idealised analytical models or detailed finite element models. Various limitations are inherently associated with each of these techniques when used in isolation, including the relevance to *in-vivo* conditions, the identification of accurate initial conditions, boundary conditions and material properties, and the errors arising from simplifying assumptions. However, when a number of these techniques are used in complement to each other, it is possible to reduce levels of uncertainty, thereby increasing the accuracy and quality of analyses. This topic has been a subject of research in University College Dublin over the past decade which has involved the use of idealised and anatomically realistic two-dimensional [1-5] and three-dimensional [6-13] finite element computational models, in-depth accident reconstruction techniques [14-19], and laboratory experimentation [20-23]. During the course of this research, we have provided free and direct access to detailed three-dimensional finite element head models [7, 10, 12] and Matlab code for automatically constructing hexahedral finite element meshes [24, 25] from sets of digitised data which is typically in the form of either CT or MRI scans, for research purposes. Such resources are available from the corresponding author upon request and are of use in impact biomechanics research [26].

Most recently, we have completed an analysis of a set of ten documented cases of real-world accidental falls [19], from which people sustained non-fatal head injuries which required a level of neurosurgical intervention. An illustrative one of these analyses is the focus of this present paper and is summarised along with a second case in Table 1. Details of the remaining cases are provided elsewhere [27]. All cases were screened to limit the selection to relatively simple falls, in order to facilitate modelling of the accidents and to avoid cases of polytrauma in which it would have been difficult to partition the proportions of energy associated with particular injuries. Only accidents resulting in focal injury were considered. Clinical assessments and accompanying CT scans of each case were provided by the National Department of Neurosurgery at Beaumont Hospital, Dublin. Each accident site was examined shortly after the accidents to determine the layout of the environment, the height of the fall, and the type of surface onto which the person fell. The engineering analysis involved reconstructing the dynamics of each fall, while carefully evaluating the associated boundary and initial conditions, and comparing each reconstruction against the clinical evidence of injury. The results of these multibody dynamics reconstructions are available in the form of time-varying profiles of linear and angular accelerations and force, and injury criteria including HIC, GAMBIT and two power indices, HIP and PI. These, in turn, can be related to thresholds for head injury that have been reported in the wider research literature. Of particular relevance to this present paper, however, are the time-varying profiles of the linear and angular velocity vectors of each accident case: those of the illustrative accident are presented in this paper in graphical format. The same data is freely available for all ten cases in electronic tabular form directly from the authors. It is hoped that this will serve to provide a valuable research resource for use by the wider research community which is concerned with head injury and the development of measures to protect against head injury. Such in-depth information and analysis could

usefully form a database of clinically relevant cases for head injury modelling by many different and independently developed finite element human head models.

Case	Sex	Age yrs	Height cm	Weight kg	Brief description of accident	Summary of head injury
1	F	76	160	60	Lost balance and fell directly backwards. Incurred occipital impact of head against concrete wall.	Small left frontal lobe contusion. Large right temporal parenchymal haemorrhage.
2	M	11	152	37	Fell directly backwards. Incurred occipital impact of head on concrete pavement.	Right frontal lobe contusion. Traumatic subarachnoid haemorrhage.
3	M	85	163	70	Fell directly forwards after losing balance. Incurred frontal impact on concrete pavement.	Left sided chronic subdural haematoma. Right sided acute subdural haematoma. Midline shift to left.
4	F	84	163	63.5	Fell directly forwards after losing balance while walking downhill. Right fronto-parietal impact on concrete footpath.	Left sided subdural haematoma with midline shift. Dilated right ventricle.
5	F	84	163	57	Tripped on a crack and fell forward and to her right. Right fronto-parietal impact off concrete footpath.	Right sided acute and chronic subdural haemorrhage with midline shift and subfalcine herniation.
6	F	71	163	63.5	Lost balance and fell forwards. Left fronto-parietal impact off the concrete ground.	Left fronto-parietal subdural haematoma with midline shift and asymmetrical ventricles.
7	M	76	170	66.7	Tripped and fell forward on right side; broke his shoulder and hit right side of his face and head.	Right chronic subdural haematoma. Left acute subdural haematoma.
8	F	87	157	51	Slipped on a ramp and fell forward and hit the right side of head off a railing.	Left sided subdural haematoma. Lateral ventricle shifted to the right side.
9	M	37	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.
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.

Table 1. Summary details of the ten accident cases and associated head injuries (Case 1 is analysed in greater detail in this present paper; the remaining cases are analysed fully in [27])

METHODS

Accident reconstruction is carried out using MADYMO (MAtheMatical DYnamic MOdels) multibody dynamics software [28]. MADYMO has a database of dummy models, which makes it very suitable for reconstructing accidents involving humans. For these analyses we use the pedestrian models, of which five were available: 95th and 50th percentile male, 5th percentile female, 6 year old child, and 3 year old child. Gravity must be included as an acceleration field, and contact specified between parts of the body and the environment. Contact force is calculated by means of a force-penetration function assigned to either or both of the contacting surfaces. Default force-penetration functions are supplied for most of the ellipsoids of the pedestrian model, and simulate impact of the ellipsoid with a rigid surface. For this analysis the head contact characteristics used by MADYMO were altered. It was found that the values for the forces and accelerations experienced by the heads of the pedestrian models were very high in comparison to values cited in the literature. This has also been found by other researchers [15, 29] who proceeded to look at the effect of using alternative head contact characteristics from the literature. From previous research [14, 15] it was

determined that the head response curve determined by Yoganandan et al [30] was the most suitable for this analysis since it was independent of the head impact location. It should be noted that the MADYMO head contact characteristics are based on tests performed on the EEVC headform. It was found that the simulations using the response curves of [30], which were obtained from experiments with actual cadaver heads, gave more realistic results in terms of accelerations and forces. Consequently, it is the head force-penetration response that is assumed in the following simulations. Figure 1 shows the difference between the MADYMO default force-penetration curves, and the average curve found by [30].

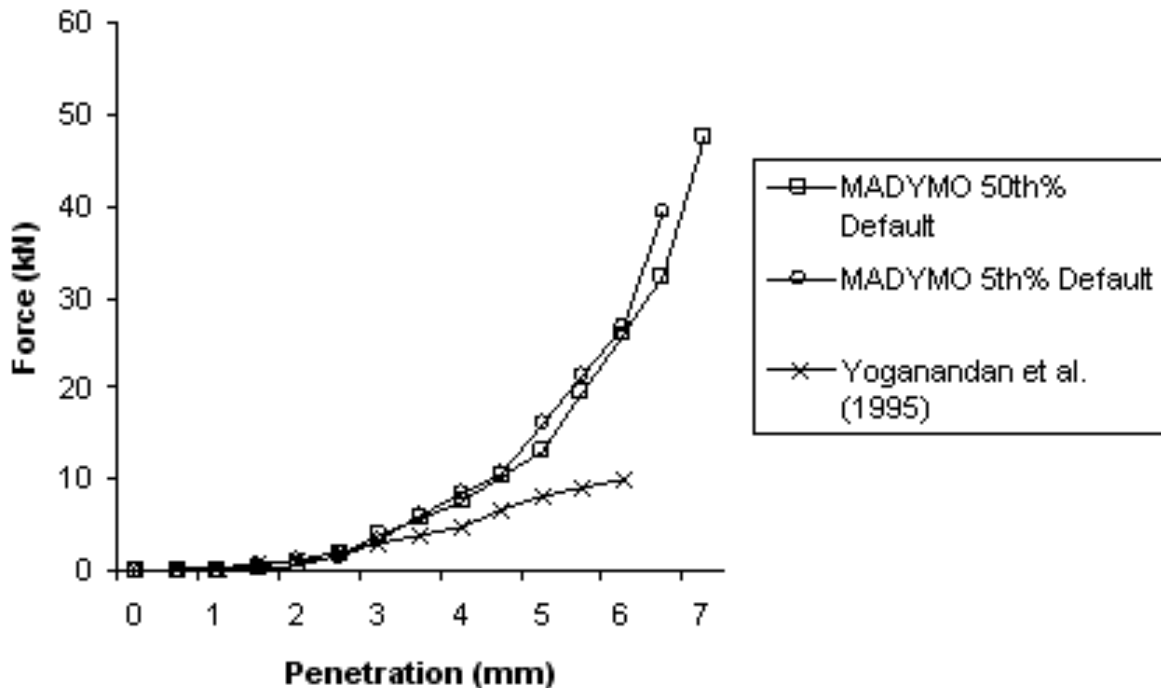


Figure 1. Contact force-penetration curves of head against ground. Yoganandan et al. [30]

Details of the various initial conditions and sensitivity analyses of the simulations are provided elsewhere by the authors [27].

EVALUATION AND ANALYSIS OF CASE 1

This case involved a 76 year old lady (height 160 cm, weight 60 kg) who lost her balance while standing on the step to the back door of her house. She was facing the door and she fell straight backwards, with the layout of the environment suggesting she hit the back of her head against a vertical concrete/cement wall which was directly behind her.

This lady sustained no loss of consciousness and presented to hospital with a GCS score of 14/15 (representing a confused state). Clinical examination of the patient revealed that the fall resulted in impact to the occipital bone, slightly to the left of midline, as evident by a 3cm laceration on the scalp overlying this region. There was no apparent skull fracture. A graze on the right elbow was also sustained, but no other external injuries were noted on full clinical trauma survey of this patient (i.e. neck, chest, abdomen, pelvis, and extremities). Computed tomography (CT) axial head scans showed a large parenchymal haemorrhage of the right temporal lobe, and a small focal bleed (contusion) on the cortical surface of the left frontal lobe.

This case was modelled using the 5th percentile female as this was closest in height and weight to the patient. Rigid planes in the inertial space were used to model the ground, the step and the concrete wall off which the lady hit her head. The pedestrian model was placed within the accident environment and initial conditions were applied to the model so it would move in accordance with the accident description.

The initial conditions used for the simulations are given in Tables 2 and 3, along with the different initial conditions used for the sensitivity analysis. For this case, no initial velocity was applied to the model. However, the sensitivity analysis looked at the effect of applying different levels of backward rotational velocity.

Y-rotation	0.0 (Datum)	-0.1	-0.5	-1.0	-5.0
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Table 2. Initial joint velocities (units = rad/s) used for the different simulations

	Datum	+10%	-10%	+50%	-50%
Human Y-axis	-0.15	-0.165	-0.135	-0.225	-0.075
Left ankle Z-axis	0.15	0.165	0.135	0.225	0.075
Right ankle Z-axis	0.15	0.165	0.135	0.225	0.075

Table 3. Different configurations used for initial orientation of joints at the beginning of the datum simulation and simulations in which values were changed by $\pm 10\%$ and by $\pm 50\%$. Values are in radians

Figure 2 shows a series of stills taken from the MADYMO simulation showing the sequence of the fall. Figure 3 shows the computed components of linear and angular velocity immediately prior to and during the sequence of the impact event. Various algorithms are available to differentiate these velocity profiles to obtain corresponding time profiles for linear acceleration and for angular acceleration; these are provided elsewhere [19, 27] by the authors. Figure 4 shows the CT scans indicating the frontal contusion and the parenchymal haemorrhage. It should be noted that the CT images are taken from the bottom up, so the left sided contusion appears to be on the right side as the observer looks at the scan.

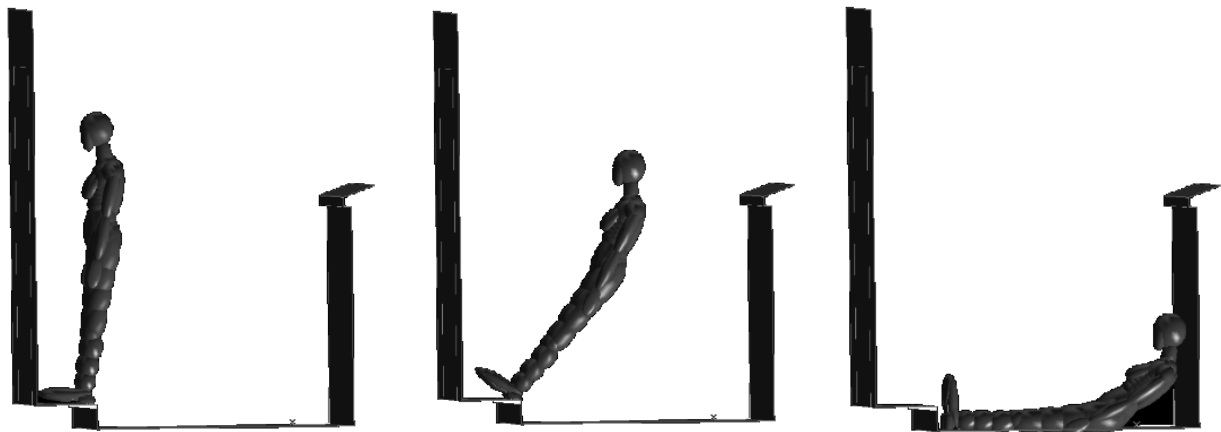


Figure 2. Still images taken from the MADYMO simulation of accident Case 1

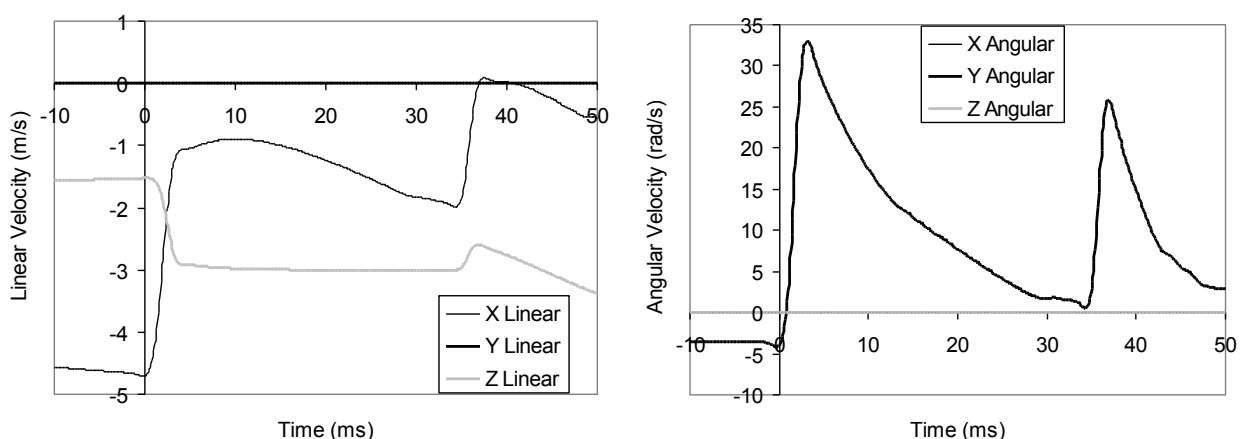


Figure 3. Components of linear and angular velocity from 10ms prior to head impact and during the impact event. All results are defined with respect to a right-handed coordinate frame in which the origin is at the head's centre of gravity and the x-axis faced forwards

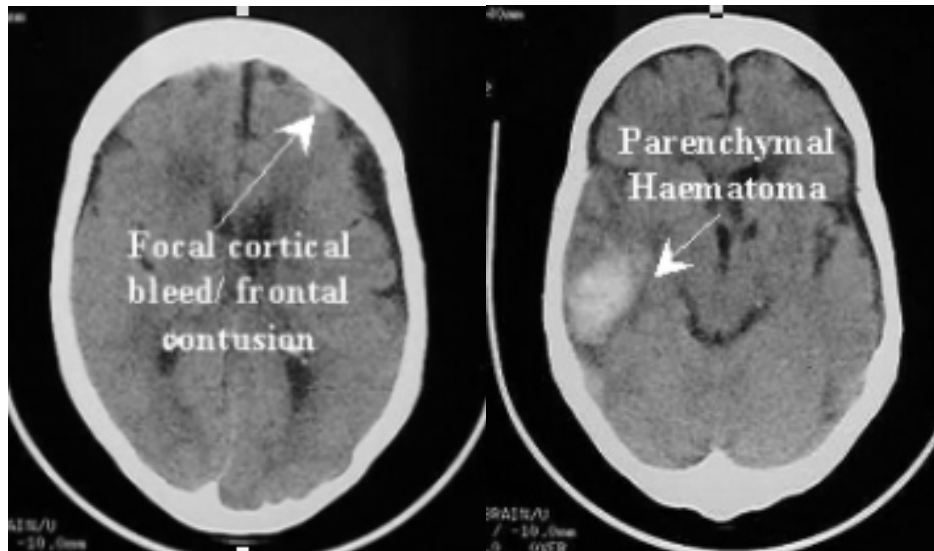


Figure 4. CT scans showing (left) the frontal contusion and (right) the parenchymal haemorrhage

DISCUSSION

The peak values of linear acceleration for the ten cases range from 236.5 to 366.5 g. The duration of the impact is approximately 3 ms for all cases due to the fact that the impact surface was generally rigid concrete. These values all fall above the tolerance proposed by [31] who stated that a deceleration of 200 g with a pulse duration of 3.5 ms would be sufficient to cause bridging vein rupture resulting in acute subdural haematoma (ASDH). The pulse duration was found to be an important factor since the rate at which the deceleration occurred affected the injury outcome. As the pulse duration was increased, with the deceleration constant, ASDH was no longer observed, thus indicating a strain rate sensitivity for this kind of injury. The values observed in the case studies here all exceed the conditions required for ASDH, and in six of the ten cases this injury was observed.

The HIC injury parameter is based solely on levels of linear acceleration over a period of either 15ms or 36ms. The HIC-36 value of this Case 1 was calculated as 780 while those of the remaining nine cases ranged up to 3419. A HIC-15 value of 700 is considered to correspond to a 5% risk of severe (AIS=4) head injury whereas a HIC-36 value of 1000 indicates a low risk of severe head injury. In nine of the ten cases (except for Case 1), the threshold of 1000 is exceeded, indicating the possibility of a severe head injury. Case 1 also sustains serious injury but this is not reflected in the HIC value. In this case there is also substantial angular acceleration present which is not taken into account by the HIC. While the HIC has been a useful parameter for designing and improving safety systems in the vehicle industry, this case highlights that it is not sufficient for cases involving rotational motion. In a study of helmeted impacts carried out by [33], they also found estimated HIC threshold levels for 25%, 50% and 80% of probability of MTBI to be 151, 240, and 369. Again this highlights the fact that HIC does not take rotational components of motion into account, and that serious injury cannot be ruled out even if HIC remains below a threshold level.

Auer et al [32] produced a series of tolerance curves relating the occurrence of subdural haematoma (SDH), subarachnoidal haematoma (SAH) and contusion to linear acceleration. For all injuries it was found that as the pulse duration decreased, the magnitude of the acceleration that could be sustained before injury increased. For all the cases in this study, the acceleration levels and pulse durations fall within the upper and lower tolerance curves for SDH and SAH, and above the lower tolerance curve for contusion. This indicates that these injuries are a possibility for all cases but not a certainty (as would be indicated if the values fell above the upper tolerance curve). Zhang et al [33] found that the maximum resultant translational acceleration of the head for a 25%, 50% and 80% of MTBI was estimated to be 66, 82 and 106 g respectively. The injury tolerance estimated was primarily derived from a typical impact duration of 10 to 16 ms. Although these values of acceleration are considerably lower than those from the case studies, the duration of the impact is much longer. Comparing the results of [32] and [33] it can be seen that similar tolerance levels are found for the longer time duration.

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 level of linear acceleration is above the lower tolerance curve for contusion proposed by [32] indicating that contusion is predicted to be 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.

Many criteria based on various mechanical metrics including linear acceleration, angular acceleration and force have been proposed to indicate traumatic brain injury. These include HIC, GAMBIT [34], Head Impact Power [34], the Power Index [35] and Peak Virtual Power [36]. While the present paper does not attempt to suggest whether any one of these criteria is more appropriate for predicting injury, further details of various such measurements are provided elsewhere [27] for each of the present ten accident cases.

CONCLUSIONS

Accident reconstruction was carried out using MADYMO multibody dynamics software. The pedestrian models were selected for accident reconstruction due to their ease of use. While these models provide useful results, it is important to be aware of some of the limitations of this software. The human body is represented by a series of rigid body ellipsoids connected by kinematics joints. Deformation of soft tissue is represented by force-penetration characteristics assigned to each ellipsoid. As mentioned previously, this characteristic was altered for the ellipsoid representing the head in this study to be in line with cadaveric data rather than the aluminium headform used in pedestrian-vehicle impact testing. This change was found to give more realistic results in terms of head acceleration. The ability to scale the pedestrian models to different heights and weights has recently become available, and the use of this feature would also lead to improvement in the results of the simulations. The existing pedestrian models were validated for whole body pedestrian tests, so it can be assumed that they provide a good representation of the human body. Due to the simplicity of the models there is scope for improved biofidelity. However, further investigations into this area was outside the scope of this study. Improved biofidelity would lead to more accurate modelling of accident cases, yielding improved 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. On the other hand, 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. The results from multibody modelling, in the form of velocities, accelerations or forces, can be used subsequently as input for three-dimensional finite element models of the head. Such analyses [16-19] give output in the form of brain tissue deformation resulting from the head impact.

Time-varying profiles of the linear and angular velocity vectors of one of ten accident cases have been presented in graphical format in this paper. The same data is freely available in electronic tabular form directly from the authors. It is hoped that the in-depth information and analysis of these ten real world accidents will constitute a valuable database of head injury cases for subsequent research by the wider head impact biomechanics community. Various three-dimensional finite element brain trauma models that have been developed within University College Dublin have been used to simulate each of these cases, and to predict the transient evolution of stresses, strains and related metrics within the neural tissue [19].

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