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RECONSTRUCTION OF HEAD INJURY CASES ARISING FROM FALLS USING THE UCD BRAIN TRAUMA MODEL

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Abstract. While Road Traffic Accidents continue to be the largest contributor to head injury, falls are usually second in prevalence. This paper looks at numerical modelling techniques, namely multibody body dynamics and finite element methods, in order to reconstruct two real-life accident cases arising from falls. Various modelling strategies are explored, and the results are compared with existing published brain injury tolerance levels.

Key words: impact biomechanics, falls, multibody dynamics, accident reconstruction, head injury, finite element modelling.

1. INTRODUCTION

Traumatic head impact injuries occur when the human skull and brain are rapidly subjected to intolerable levels of energy. There exist many causes of neurotrauma: accidents, falls, assaults and injuries occurring during occupational, recreational and sporting activities. While Road Traffic Accidents tend to be the leading cause of injury related death, falls tend to be the leading cause of non-fatal hospitalisation [1, 2]. 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 fall admissions [2].

The aim of this research is to reconstruct a number of real life falling accidents using numerical techniques in order to establish injury criteria for specific types of brain lesion. Firstly, the accidents are modeled using multibody dynamics software to recreate the overall movement of the body during the accident. Next, the output from the multibody dynamics simulation is then used as input for the University College Dublin Brain Trauma Model (UCDBTM) [3]. The UCDBTM 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 clinical results observed.

2. METHODOLOGY

The head injury research group at University College Dublin have been working in conjunction with the National Department of Neurosurgery at Beaumont Hospital, Dublin, obtaining head injury cases arising from simple falls. Using information from witness accounts and neurosurgery reports, two such cases were reconstructed using numerical methods [4].

Case 1: Boy fainting at a fountain.

Suffering from heat exhaustion, an 11 year old boy (height 152 cm, weight 37 kg) fainted after straightening up from drinking at a water fountain on a city street. According to witnesses he fell straight backwards and his head rebounded off the ground. The ground was reported to be level and concrete. The patient experienced a brief loss of consciousness. Upon revival his Glasgow Coma Scale (GCS) score was 14/15, indicating mild confusion. Detailed clinical examination of the patient revealed that the fall resulted in impact just over the occiput, mostly central, and slightly right-sided, as evidenced by subcutaneous bruising and swelling of the scalp overlying this region. There was no apparent skull fracture and no other injuries were noted on full clinical trauma survey of this patient (i.e. neck, chest, abdomen, pelvis, and extremities). CT imaging on presentation to hospital revealed a right lateral frontal intracerebral haemorrhagic contusion. There was also evident blood in the right Sylvian fissure seen on CT imaging, representing traumatic subarachnoid haemorrhage.

Case 2: Lady falling from step.

The second case involved a 76 year old lady (height 160cm, weight 60kg) who lost her balance while standing on the back step of her house, facing the door. She fell straight backwards, with the layout of the environment suggesting that she hit her head off a vertical concrete/cement wall. This lady did not lose consciousness and presented to hospital with a GCS score of 14/15 (representing a mildly confused state). Detailed clinical examination of the patient revealed that the fall resulted in impact to the occipital bone, slightly to the left of midline, and this was 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. CT scans revealed a large parenchymal haemorrhage of the right temporal lobe, and a small focal bleed on the cortical surface of the left frontal lobe.

MADYMO [5] multibody dynamics software was used to reconstruct the falls. MADYMO contains a suite of pedestrian models and these cases were modeled using the 5th percentile female model, since this was closest in height and weight to both patients. The features of the environment that may have affected the fall were modeled using planes and cylinders. For both of these simulations, the model was positioned in the inertial space leaning slightly backwards (backward rotation of 0.1 radian) and allowed to fall with gravity without any initial velocity being applied. Contact was specified between the environment and the ellipsoids of the body relevant for each fall. The force-penetration characteristics of the head were altered to represent more realistic contact. An analysis of the sensitivity of the results to changes in initial conditions was carried out and the range of results was sufficiently small to not require multiple FE simulations to be run [4].

The baseline UCDBTM, which was refined to differentiate between the grey, white and ventricular matter, was used for this analysis. The scalp elements (which completely encased the head apart from the foramen region where the brain elements are unconstrained) were redefined to be rigid and the full time histories of the six velocity components of the centre of the MADYMO head were applied to a reference node located at the COG of the head. The FE head model was also scaled so as to represent the same weight head, which was used in the multibody simulations, since this affects the intracranial pressure, Von-Mises stress and shear stress responses [3].

Applying the full kinematic time-histories to the model required excessive CPU time with the current hardware resources (up to 17 days to solve) and consequently two other modeling strategies were explored in an effort to reduce the time required to simulate the impact. For studying brain injury mechanics it is the time during the impact event that is most important. Two methods were devised by which the simulation time could be reduced. In both cases the analysis was run from a time closer to when the impact occurred. The first method, referred to as "abrupt run", involved applying the velocity profiles at a short time before impact directly to the head at the start of the analysis. The head was orientated at the appropriate angle relative to the ground before the simulation began. This is a very abrupt method of starting the simulation, however it was established that there was enough time before the impact occurred for the starting effects to attenuate (by examining the evolving stresses in various elements). In the second approach, which is referred to as the “adjusted run”, the velocities before impact were taken as the datum, and the velocities relative to these were applied to the rotated head. This meant that the analysis effectively started from stationary and the movement was applied more gently to the model. The pressure, Von-Mises response, shear stress response and logarithmic strain values were compared for all three analysis strategies in the frontal, parietal, occipital and midbrain regions. As seen in Figure 1 the shortened simulation strategies were identical to the results of the full simulation and thus the “adjusted run” method was chosen as the strategy for modelling further accident reconstructions: this method required the least computation time (typically only three days).

Having chosen this method, the FE model was tested for both cases, comparing the coupled node brain/CSF approach with a sliding brain/skull boundary approach [6]. In this second approach the solid elements previously defined as CSF material were redefined to correspond to the material of the dura, falx and tentorium (the shell elements which had previously been assigned to these regions were removed from the model). The simulation therefore modelled the brain in direct contact (sliding with friction) with the intracranial membranes and the inner surface of the skull.

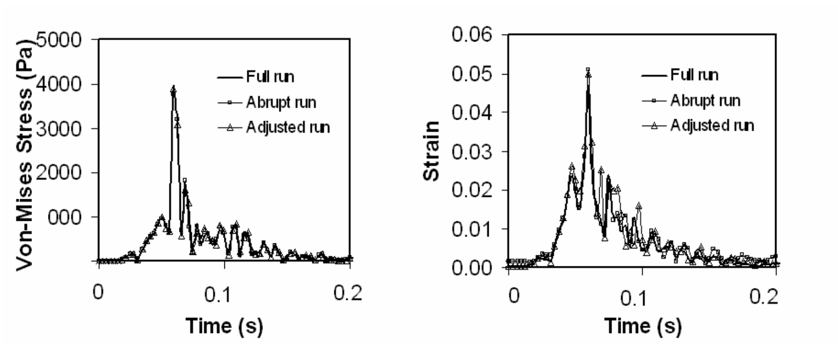

Figure 1. Comparison of methods for reconstructing case studies. These graphs are representative of the curves observed for all regions analysed. It is evident that the faster abrupt and adjusted modeling strategies work sufficiently.

The results of the multibody dynamics simulations are presented in Table 1 along with a summary of the relevant literature. A similar table for the finite element simulations is presented in Table 2, which reports the levels predicted in the area where the actual injury occurred as well as the maximum values predicted for the finite element brain.

Table 1. Table of outputs from MADYMO simulation for both cases.

Output Parameter	Simulation Value		Comments
	Boy	Lady	
Force Duration	12.1 kN < 2 ms	5.134 kN < 1.5 ms	8.8 – 14.1 kN → Fracture [7] 5.8 – 17 kN–→ Fracture [8]
Resultant Linear Velocity (m/s)	6.88	3.77	
Resultant Angular Velocity (rad/s)	30.21	25	Below gliding contusion threshold [9]. Inside bounds for DAI and concussion to be absent [10].
Resultant Linear Acceleration (g)	584	308	Above lower bound for contrecoup contusion, SAH and SDH [11]. Above 100% concussion probability [12].
Resultant Angular Acceleration (rad/s^2)	20700	44400	Below tolerance curve for SDH [13]. Above sustainable levels observed [14]. Above 100% injury probability of MTBI [12].
HIC	8690	1150	Boy: Level indicates fatal injury sustained. Lady: Above 50% injury probability [15]
GAMBIT [16]	3.03	5.36	>1 is injurious
HIP (kW) [12] 50^{th} %ile values: Scaled FE head*:	 92.3 66.7	 21.5 17.5	Head Impact Power - 50% injury probability at 12.8kW.
PI (kW) [12, 17] 50^{th} %ile values: Scaled FE head*:	 118 87.5	 27 23	Threshold for power index subdural haematoma for posterior anterior rotation of 50kW.

*Mass: 3.136, Ixx: 0.01131, Iyy: 0.01102, Izz: 0.007714

Table 2. FE simulation results for both cases. Methods 1 and 2 refer to models using the tied CSF definition and the sliding brain dura boundary approach [6] respectively.

Parameter	Method	Simulated Value				Literature Value
		Injury area		Highest Value *		
		Boy	Lady	Boy	Lady	
Von Mises Stress (kPa)	1: 2:	3.2 3.26	6 5.09	12.67 9.64	18.8 15.17	15-20 → concussion [18,19] 30 → AIS 2+ [20] 7kPa for approach A, 8.6kPa for approach B → contusion [6]**
Strain	1: 2:	0.07 0.09	0.2 0.17	0.3 0.27	0.28 0.43	>0.25 → structural failure [21] >0.20 → functional deficit [21] <0.10 → reversible injury [21] 0.188 → BBB injury [22] 9% for approach A, 7.6% for approach B → contusion [6]
Strain Rate (s^{-1})	1: 2:	11.8 9.36	30 26.1	71.63 38.2	57.7 50	23-140 → injury [23] 11-67 → non injury [23]
Strain x Strain Rate (s^{-1})	1: 2:	0.48 0.14	2.4 0.62	19.5 2.4	20 4.44	10-80 → injury (concussion) [23] 0-20 → no injury [23]

* Maximum values achieved at different locations for both methods.
** Approach A is Miller *et al* version of Method 1, Approach B is similar to Method 2.

Figure 2 shows a cross section of the FE brain showing the state of maximum strain for Case1. Figure 3 shows a cross section of the FE brain showing the state of maximum Von-Mises stress for Case 2. For both simulations the differences between the two simulation approaches are evident. In both cases the highest value reported by the sliding model is in closer proximity to the location of the actual injury. There may be an unrealistic constraint put on the motion of the brain under the current configuration used for the CSF in the coupled node model.

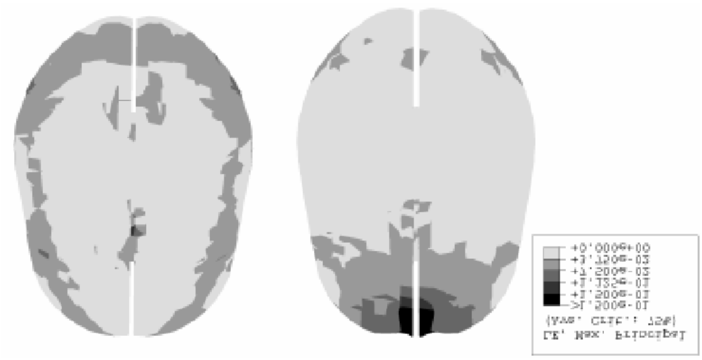

Figure 2. Comparison of strain results using the two modelling methods for Case 1. On the left is the fixed node definition and the right the sliding definition. An obvious higher strain exists in the frontal lobe area of the sliding model. This is also where contusion occurred.

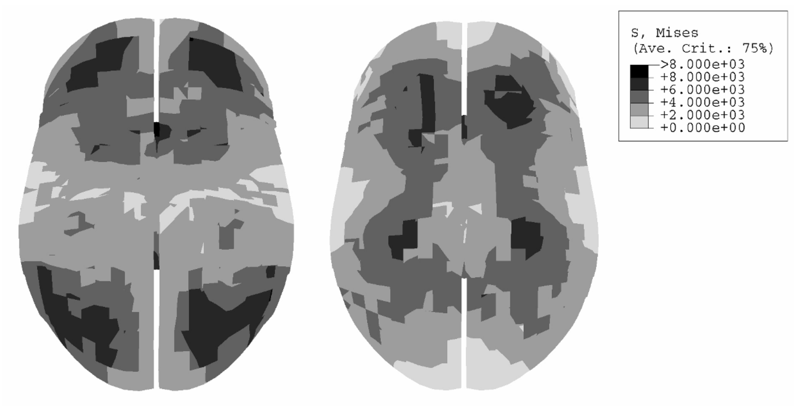

Figure 3. Off centre sagittal view of the Von-Mises stress in the FE head during Case 2. The left image shows the prediction of the tied boundary model, while the right is that of the sliding boundary model. The sliding boundary model shows stresses set deeper into the brain, with the coupled node model experiencing peaks nearer the cerebrum's periphery. Again the sliding model correlates better with injured regions.

3. DISCUSSION

The results from the two cases are considerably different, especially when looking at the results of the MADYMO simulations. Some of the results observed for the fountain case fall well above the various threshold values cited in literature, indicating that the boy should have sustained much more serious injury than that observed clinically. The case of the lady on the step gives much more realistic results in terms of HIC and linear acceleration, however angular acceleration values are very high here. The oblique nature of the impact is more likely to result in higher angular accelerations than linear impact would.

The MADYMO model used, while being the 5th percentile, was still heavier than the actual boy. Mohan *et al.* [24] suggest that at age 13, skull stiffness is approximately 90% of adult skull stiffness. This boy was 11, and it is possible that his skull is not fully calcified compared to an adult, leading to lower head stiffness. The joint stiffnesses of the model were not analysed for the purpose of this study. Tolerance limits are also scarce for children. If these factors could be represented more accurately, the linear and angular accelerations would be more accurate, and hence more insight into injury of children may be found from the FE model.

In the case of falls onto rigid surfaces the impact duration is usually quite short with an associated high peak value. Even though very high values for acceleration are observed in these cases, the duration for which these peaks occur is very short. This factor is not taken into account by injury measures such as GAMBIT.

The results of the finite element analysis have shown that the UCDBTM demonstrates stress and strain levels in agreement with those of the literature, though availability of tolerance limits for fall cases are fewer than for those obtained from tests conducted for investigating RTAs. For the study of contusion, the case studies suggest that a sliding boundary model has better prediction capabilities, with the sliding model showing higher strain and Von-Mises stress levels in closer proximities to the region that was injured. For the case studies analysed, the tied brain/CSF model always experienced maximum strain and stress in the midbrain region. The tied model also experienced higher levels of stress and strain in the occipital lobe than the sliding model, but no injury was observed in that location.

4. CONCLUSIONS

Due to the large amount of CPU time required to run a finite element analysis of a head impact resulting from a simple fall, other modelling strategies were explored in an effort to reduce this time. It was found that two alternative strategies produced very similar results, and so it was decided that the "adjusted run" method be used for future analyses since this required least CPU time.

In order for the prediction capabilities of the reconstructions to become more reliable improvements would be required for the UCDBTM. These include increasing the mesh density of elements next to the fluid elements and an improved representation of the boundary region of the ventricles. The MADYMO reconstruction of the case studies presented here suffered high levels of angular and linear acceleration over very short durations, with their values usually exceeding what is generally accepted as injurious.

The results observed for these two cases suggest that this method will be useful for investigations into tolerance limits acceptable to the human for different types of injury, provided correct modelling methods are used. Future work involves the analysis of more accident cases in the hope of finding tolerance levels of brain deformation which result in specific brain injuries.

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