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Authors(s)	O'Riordain, K., Thomas, P. M., Phillips, J. P., et al.
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RECONSTRUCTION OF REAL WORLD HEAD INJURY ACCIDENTS RESULTING FROM FALLS USING MULTIBODY DYNAMICS

K. O’Riordain^a, P.M. Thomas^b, J.P. Phillips^b, M.D. Gilchrist^{a,*}

^aDepartment of Mechanical Engineering, University College Dublin, Belfield, Dublin 4, Ireland.

^bNational Department of Neurosurgery, Beaumont Hospital, Dublin 9, Ireland.

* Corresponding author: michael.gilchrist@ucd.ie

Abstract

Objective To reconstruct real life head injury accidents resulting from falls using multibody modelling software, with the aim of comparing simulation output to injuries sustained.

Background Much previous research on head injury biomechanics has focussed on animals and cadavers. However, focus is increasingly turning towards the examination of real life head injury. Falls are a major cause of head injury and, in general, are simpler to model than other accident types.

Design and Methods Five cases of simple falling accidents resulting in focal head injury were examined, and reconstructions were performed using a multibody model of the human body. Each case was reconstructed a number of times, varying the initial conditions and using two different sets of properties for head contact.

Results Results obtained included velocities, accelerations and forces on the head during impact. This output appeared more sensitive to changes in head contact characteristics than to changes in initial conditions. Depending on the contact characteristics used, results were consistent with proposed tolerance limits from the literature for various lesion types.

Conclusions Provided it is used with caution, this method could prove a useful source of biomechanical data for the investigation of head injury biomechanics.

Relevance

Biomechanical investigation of real-life cases of head injury is very important, yet not as prevalent as work with animals and cadavers. Reconstruction of real life accidents is a good method of obtaining data that will aid in the investigation of mechanisms of head injury and human tolerance to head injury.

Keywords: Head injury; Brain; Multibody modelling; Falls

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]. Head impact injuries account for approximately half of all deaths due to mechanical trauma, but account for the majority of cases of disability after injury. In addition to the huge human cost to society, the financial cost of hospitalisation, care and rehabilitation of head injured people has been estimated to be as high as \$33 billion per year in the US alone [2]. Much research has focussed in the past few decades on the biomechanics of traumatic head injury - attempting to discover the exact mechanisms of different types of head injury and the tolerance limits of the head and brain to impact. The eventual aim of this research is to reduce or avoid damage by reducing or eliminating the mechanism that causes the damage. Simulation tools offer potential to design appropriate helmets and protective devices for the human body.

1.1. Epidemiology

Road traffic accidents (RTAs), falls and assaults are the most frequently cited causes of head injury. However, the ratios of these three differ worldwide, as summarised in Figure 1. While RTAs tend to be the leading cause of injury related death, falls tend to be the leading cause of non-fatal hospitalisation [6-7]. 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 [7]. It is also important to note that RTAs and falls generally lead to different types of trauma. While RTAs tend to cause diffuse and multifocal injuries, the type of head injuries received from falls tend to be focal. Contrecoup injury, where the injury occurs at the contralateral side to the site of impact, is particularly prevalent with falls, while ipsilateral (or coup) injury rarely occurs except in cases involving skull fracture [8-9]. A study examining the pattern of brain injuries in falls resulting in death [9] found skull fracture in 66% of cases. This study found that in almost all cases of skull fracture there was both coup and contrecoup injury. However, others [8] have found that the occurrence of skull fracture decreases the likelihood of contrecoup injury. Acute subdural haematoma (ASDH) have been found in between 78% and 85% of fatal falls [9-10]. Indeed, Gennarelli and Thibault [11] suggest that 72% of ASDHs are due to falls or assaults. Zwimpfer *et al.* [12] found that almost all patients admitted for head injury resulting from falls due to seizures had intracranial mass lesions. Of these, 85% were extra-axial lesions - either extradural or subdural haematomas.

1.2. Head Injury Biomechanics

Research into the biomechanics of traumatic head injury concentrates on four main areas, namely mechanisms of injury, response of brain tissue to impact, tolerance limits of tissue to injury and the development of injury assessment technology. Historically, much of the research on the biomechanics of traumatic head injury has been conducted on animals and cadavers. Indeed, the majority of literature on tolerance limits of the head to impact, and the development of severity indices for head impact, have resulted from this type of study.

The Head Injury Criterion (HIC) [13], for example, is the most widely used head injury severity index, originally developed from cadaver and animal studies. It is a predictor of the risk of sustaining head injury, calculated from linear acceleration of the head:

$$HIC = \max_{T_0 \leq t_1 \leq t_2 \leq TE} \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} R(t) dt \right]^{2.5} (t_2 - t_1) \quad (1)$$

where T_0 is the starting time of the simulation, TE is the end time, $R(t)$ is the resultant head acceleration (measured in g) over the time interval $T_0 \leq t \leq TE$, and t_1 and t_2 are the initial and final times (in seconds) of the interval during which the HIC attains a maximum value. A HIC value of more than 1000 is taken to represent an unacceptable risk of injury. While this is the current standard in the automotive industry, there are many critics of the HIC who argue, among other things, that a single criterion based on a single parameter should not be used to describe tolerance to all kinds of head injuries for all parts of the population. More recently, there has been a move away from this approach of looking for a parameter that correlates well with overall severity of injury, and many are now focussing on determining tolerance limits of the head to specific lesion types, for example, acute subdural haematoma (ASDH), diffuse axonal injury (DAI) or skull fracture.

For example, Gennarelli and Thibault [11] produced ASDH experimentally in monkeys, and comparing the results with an analytical model and human clinical data, proposed tolerance curves for ASDH due to rupture of bridging veins. They suggest that bridging veins are highly sensitive to strain-rate and tend to rupture during impacts associated with high rates of increasing acceleration. As the duration of the pulse increases, higher levels of angular acceleration will be required in order to maintain the high strain rate necessary for rupture of bridging veins. Figure 2 shows their tolerance curves for rhesus monkeys. For humans, they suggest that a fall resulting in head acceleration of over 200 g and a pulse duration of 3.5 ms or less would create conditions necessary for the production of bridging vein ASDH.

Margulies and Thibault [14], investigating diffuse brain injury, proposed a human injury tolerance curve for DAI and milder forms of axonal injury such as cerebral concussion. Again, animal studies and analytical models were used to determine critical levels of strain associated with the onset of DAI and concussion. Rotational motion was an appropriate indicator of this critical strain, in particular peak change in rotational velocity and peak rotational acceleration (i.e., the difference between the maximum and minimum rotational velocities and rotational accelerations). Moreover, the particular plane in which this rotational motion occurred was also of significance. It was hypothesised that the importance of each of these parameters in causing diffuse injury alters across different impact conditions. Figure 3 shows the tolerance curves (as reproduced by Ryan and Vilenius [15]) derived for 5% critical strain, below which there should be no axonal injury, and 10% critical strain, below which mild injury such as concussion could be expected and above which DAI can be expected. These authors suggest that for impacts with very stiff contacts and short durations, the brain will move relative to the skull at impact, and thus a change in angular velocity of the skull will be of prime importance in causation of injury. However, for impacts with softer structures, the brain will tend to move with the head, and will thus be subjected to the same accelerations. It is only in this latter case that diffuse injury occurs.

Tolerance of the head to skull fracture is much easier to determine than tolerance to intracranial injury. This is because of the definite relationship between force applied to the skull, and failure of cranial bone. Yoganandan *et al.* [16] investigated levels of force required to fracture the skulls of 12 unembalmed cadavers. Using a hemispherical impactor with a 48 mm radius, they carried out impacts to various locations on the skull at a rate of 7.1 to 8.0 m/s. Failure loads ranged between 8.8 kN and 14.1 kN, with an average of 11.9 kN. Allsop *et al.* [17], carried out temporo-parietal impacts on 31 unembalmed cadaver heads with two types of flat

rigid impactors – one circular and 2.54 cm in diameter, the other a rectangular plate 5 x 10 cm. Fracture force for the small circular plate ranged between 2.5 and 10.0 kN with an average of 5.2 kN. Fracture force for the rectangular plate ranged between 5.8 and 17.0 kN with an average of 12.4 kN. The authors concluded that there is a significant relationship between contact area and fracture force. Thus, impacts with the ground are likely to require higher forces than with a smaller impactor.

While both animal and cadaveric studies have their advantages, the most important of which is that they can be subjected to rigorous experimentation, issues are often raised as to whether they can accurately represent the processes of human head injury. Recently, efforts have been made to examine head injury patterns resulting from real life accidents rather than animals or cadavers. While there is a degree of uncertainty regarding the exact circumstances of a real life accident, particularly where an eye-witness cannot be relied upon, nonetheless, the value of obtaining information from *real life* cases of head injury is very important. This type of research has focussed on two main areas, namely the examination of the pattern of brain injury in fatalities and/or the reconstruction of real-life accidents using either mechanical or computational models of the human body or head. Indeed, a particular area of head injury biomechanics that has enjoyed a huge increase in popularity over the past few decades is the development of computational methods for creating models that can simulate motion of the human body, or parts of the human body. These have been of use in the examination of the dynamic response of the human body under various loading conditions, and thus in attempting to elucidate mechanisms of injury and tolerance of the body to injury.

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 [18] [19] [20]. Others have sought to establish tolerance limits to specific types of head injury by reconstructing accidents and comparing the injuries sustained with parameters calculated from the reconstructions. For example, Auer *et al.* [21] reconstructed 25 fatal pedestrian accidents using various methods, including computer simulation. Head acceleration and impact duration were calculated, and from these, the upper tolerance limit (lowest level of loading above which the specific injury is always observed) and the lower tolerance limit (highest value below which the injury never occurs) for various kinds of brain injury were calculated (Figure 4). The types of lesions examined were subdural haematoma, subarachnoid haematoma and brain contusions. While the authors did not elaborate on the relationship between the mechanical parameters and the lesions observed, they concluded that reconstructing pedestrian accidents could be a useful means of estimating tolerance limits for discrete brain injuries. However, due to lack of certainty about input variables, these are still very approximate estimations. Clearly, traumatic lesions can be caused by either linear or rotational accelerations of the head. Gennarelli and Thibault (Figure 2) have shown how angular acceleration will induce ASDH whilst the same type of injury has been shown by Auer *et al.* (Figure 4) to occur under the action of linear acceleration. What is currently unknown is the relative degree to which linear and angular accelerations influence the occurrence of different traumatic lesions.

1.3. Aim of present research

The aim of this research is to use multibody dynamic simulations to reconstruct real life head injury accidents resulting from falls. The advantage of this approach over previous work

described in Section 1.2 is that all simulations are of living humans who have been involved in non-fatal accidents. Multibody dynamic simulation can easily be applied to reconstruct simple accidents such as falls to provide quantitative insight into the occurrence of specific clinical trauma. Falls were selected as the accident type of interest due to the relative certainty regarding initial circumstances compared to RTAs, for example. Reconstructions were performed using a multibody model of the human body, which provided such output as linear and angular accelerations of the head during impact, and force on the head. Clinical data, including computed tomography (CT) scans, were examined and results obtained from the simulations were compared to both the injuries sustained and the literature available on tolerance to various types of head injury.

2. Methodology

2.1. Software

In the last two decades a great deal of research has led to the development of several general-purpose computer programs which simulate complex multibody systems. These methods are computationally favourable to other methods for simulating occupant and pedestrian dynamics in vehicle crashes and other accidents. A multibody system is a collection of rigid and flexible bodies joined together by kinematic joints (e.g., revolute or translational joints) and force elements (e.g., springs and dampers). It is possible to construct a multibody model of the human body with these kinematic joints and various discrete bodies of particular size and shape. The presence of these kinematic joints is conveniently accounted for in such a model by means of global and local coordinate reference systems (global would refer to the Cartesian coordinates of an initial reference frame while local would identify coordinates specific to an individual structural member such as the lower limb of a human).

MADYMO (Mathematical Dynamic Models) [22] is such a software tool that has been developed to design and optimise automotive occupant safety systems, but which has a range of other related applications (accident reconstruction, injury biomechanics, vehicle handling, etc.). It offers the capability of combining both multibody and finite element modelling techniques. The multibody approach uses numerical algorithms to predict the motion of systems of bodies connected by kinematic joints, based on initial conditions and the inertial properties of the bodies. The finite element capability allows simulation of structural behaviour, by discretising a continuum and predicting the associated constitutive response.

A particular strength of this software is its database of human body models, including the family of multibody pedestrian models. They are available in five sizes: 50th percentile (average male: weight 75.7 kg, height 174 cm), 95th percentile (large male: weight 101.1 kg, height 191 cm), 5th percentile (small female: weight 49.8 kg, height 153 cm), 6 year old child (weight 23 kg, height 117 cm), and 3 year old child (weight 14.5 kg, height 95 cm). MADYMO's pedestrian model was originally adapted from the standing version of the Hybrid III crash test dummy, altering the joint stiffnesses of the lower extremities, neck and spine and contact characteristics of the pelvis, abdomen, ribs, shoulders and head in order to make it respond in a more realistically human manner. Head contact characteristics are based on the EEVC (European Experimental Vehicles Committee) adult headform model. The EEVC proposed in 1994 [23] a

standard set of tests using various impactors that were designed to assess the potential risk to pedestrians from vehicle impacts.

The MADYMO pedestrian models have a ‘skeleton’ consisting of chains of 52 bodies. The outer geometry of the pedestrian model is based on the RAMSIS software [24] and is represented using 64 ellipsoids (66 for the female). Deformation of soft tissue (flesh and skin) is represented by force-penetration characteristics assigned to each ellipsoid. These characteristics describe contact interactions of the pedestrian model with itself and with its environment.

2.2. Data Collection

Four cases of falling accidents resulting in focal head injury were selected by the National Department of Neurosurgery at Beaumont Hospital, Dublin. Only relatively simple falls were considered, in order to facilitate the simulation. In all cases, medical examination of study individuals were carried out within 6 hours of injury, including head computed tomography (CT). While the extent of the initial injury may evolve after 6 hours, evidence of all the lesions of interest to this study (i.e., not DAI) would be apparent on CT imaging within 6 hours of the accident occurring. A collateral history of the accident sequence was also taken from at least one eye-witness. Each accident site was then examined in order to determine the layout of the environment, height of the fall, and the type of surface on which the person fell. In each of the cases presented here, impact occurred against rigid planar surfaces such as concrete or cement.

2.3. Modelling

Each accident environment was recreated as accurately as possible using ellipsoids, cylinders, planes, and facet surfaces. The most appropriately sized pedestrian model (i.e., closest in both weight and height to the person involved) was then positioned within the environment, moving its joints to recreate the stance of the person just before falling. Initial velocities of the whole body and specific joints were estimated in order to simulate the motion of the person at the time of falling. Each accident was simulated a number of times using a variety of initial velocities that would account for any uncertainty in the eye-witness estimates of initial conditions. The estimates of both angular and linear initial velocities were altered to run two further simulations, the first using initial velocities obtained by reducing each non-zero component of the linear velocity by 0.1 m/s and each non-zero component of the angular velocity by 0.1 rad/s, and the second by increasing the initial velocities by the same amounts. These increments of initial velocity were essentially selected subjectively. However, attempts to alter the initial values by greater amounts often led to inaccurate and physically unrealistic representations of the sequence of the accidents.

Each case was also simulated using alternative head contact characteristics obtained from the literature. Contact between two surfaces in MADYMO is calculated by means of a force versus penetration function. The default head contact properties for the MADYMO pedestrian model are based on drop tests using the EEVC headform [23]. Other authors, however, have obtained alternative force-deflection characteristics for the head from cadaveric head impacts. Yoganandan *et al.* [16] provide force/deflection curves for unembalmed cadaver heads under dynamic loading. Force/deflection responses from these impacts were recorded at various

locations on the cranium (the vertex, frontal bone, parietal bone, occipital bone and temporal bone) and results were plotted to produce an average force-deflection curve. Figure 5 shows the difference between the MADYMO default force-penetration curves, and the average curve found by Yoganandan *et al.*

Thus for each case, a total of six simulations were run, firstly using the default force-penetration curves for head contact, and varying the initial conditions and then using the alternative force-penetration curves, and varying the initial conditions in the same way.

2.4. Case Studies

Four cases involving falls resulting in focal head injury were simulated. Table 1 summarises each accident, providing the age and sex of the person involved, a brief description of the accident and injuries sustained. Figure 6 depicts the initial positions of each of the cases described in Table 1 whilst Figure 7 indicates the 6 hour CT scans of each corresponding person.

3. Results

The results obtained from simulations of the above cases are summarised. Table 2 presents results from simulations using the default head contact characteristics. For each parameter, a range of peak values is given encompassing the three simulations that alter the initial velocities. Table 3 presents the range of peak values for simulations using the alternative head contact characteristics.

Figures 8-10 below show time histories for linear acceleration, angular acceleration and force for all five cases. Figures 8(a), 9(a) and 10(a) show time histories from the simulations using the default head contact characteristics, whilst figures 8(b), 9(b) and 10(b) show time histories from the simulations using the alternative head contact characteristics.

There are three curves for each case, representing the three variations of initial conditions. It will be noted, however, that for Case 4 there are six curves. The height and weight of the lady involved in this particular accident were essentially midway between the 50th and 5th percentile models. To determine the effect of running the same simulation with different models, this case was reconstructed using both the 50th and 5th percentile models.

4. Discussion

The discussion will look at four main points: (1) the effect of altering head contact characteristics; (2) the effect of altering initial velocities; (3) examining results from all cases compared to each other; (4) comparing results of these simulations to tolerance limits in the literature for various forms of head injury.

4.1. Altering head contact characteristics

In the cases presented here, almost all results obtained from using the default force-penetration characteristics for head contact are extremely high, and essentially predict severe life-threatening injuries in all cases. In reality, this was not the case. Linear accelerations

calculated for these simulations are comparable to (or even higher than) those found in studies relating to free falls from heights of up to 9 metres onto rigid surfaces [9,20] or from pedestrian impacts with vehicles travelling at over approximately 50k/h [18]. This should not be the case, as the simulations in this study are reconstructions of simple falls, mostly from standing heights. Angular accelerations (ranging between 26 and 64 krad/s²) exceed almost all values reported in the literature (e.g. Thomson *et al.* [25] found angular acceleration values of between 7 and 30 krad/s² in reconstructions of severe head injury accidents involving vehicles). Force values for all of these simulations apart from Case 1 also exceed the highest tolerance limits established for skull fracture (14.1 kN [16] or 17.0 kN [17]), with fracture actually only occurring in two cases. In many simulations, these limits are exceeded by a great amount - enough to suggest much more extensive fracture of the skull than the relatively simple linear fractures present in the two cases showing skull fracture.

Results from the simulations using the alternative head contact characteristics, however, are much more consistent with the literature (for example Auer *et al.*'s [21] tolerance curves for ASDH, subarachnoid haematoma and brain contusion). Peak linear accelerations remain high, but not unreasonably so, considering all impacts occur against surfaces that are essentially rigid compared to the head. They are also not excessively high considering that all of these people sustained moderately severe head injury. Forces on the head, similarly, fall within more acceptable levels compared to the literature, and the only case that still exceeds the highest fracture limit found by Yoganandan *et al.* [16] and Allsop *et al.* [17] (Case 3) is one that actually did have skull fracture. With regards to angular acceleration figures for these simulations, they do reduce significantly, but not by the same factor that linear acceleration does.

The significant reduction in peak values from the default simulations to the alternative simulations raises the issue of the need to use accurate figures for force-deflection characteristics of the head. Evident from all the cases presented is the fact that altering the head contact force-penetration characteristics for the multibody model has a dramatic effect on output results. This is in line with the observations of Thomson *et al.* [25] that HIC values can increase by over 500% when Hybrid III properties are used to model head contact compared to cadaveric head contact properties from the literature. The MADYMO default curves are based on tests performed on the EEVC headform, while the alternative values are based on impacts carried out by Yoganandan *et al.* [16] using unembalmed cadaver heads. Yoganandan's force-deflection characteristics are consistent with those estimated by Allsop *et al.* (17) and Manavais *et al.* (9) for skull bone – ground impact interactions. Thus it is reasonable to suggest that these figures are more realistic. It is important to remember that any model of the human body or specific body part is only as good as the constitutive mechanical data upon which it is based.

4.2. Altering initial velocities

The initial linear and angular velocities of each of these falls was relatively easy to estimate since the falls were essentially due to a loss of balance from a quasi-stationary position. It would be more difficult to estimate the initial conditions of people who sustained a fall from a bicycle, motorbike or moving vehicle. While the simulations presented in the work detail the consequences of variations of ± 0.1 m/s and ± 0.1 rad/s in linear and angular velocities, respectively, a greater range was also considered (up to $\pm 100\%$). Such different initial conditions failed to predict physically realistic kinematic motions in the various fall sequences. Consequently, it can be inferred that the actual initial conditions associated with the four falls

were most likely within the range of ± 0.1 m/s and ± 0.1 rad/s of the simulations presented in the previous sections. This cannot be stated conclusively because MADYMO does not automatically account for voluntary muscle activation and this would, of course, alter the motion sequence during a fall.

In general, changing the initial velocities of the model leads to smaller variations in peak results, particularly when compared to the variations which arose from changing the head contact characteristics. However, it does produce some variation. In general, the pattern of results appear to show that increasing initial velocities in the accident reconstruction reduces the resulting linear accelerations and forces on the head. This may seem counter intuitive, but appears to be the case due to other body parts absorbing more of the force under these conditions. For example, in Case 3 increasing initial velocities leads to increased rotation of the man's body before hitting the ground with the result that the arm and shoulder absorb a greater amount of force before the head impacts the ground. This does not appear to be the case with angular acceleration and there is much more variability in the angular acceleration values resulting from the differing initial conditions. There does not appear to be any trend relating angular acceleration to initial velocities.

The actual amount by which these peak results differ as a result of changing the initial velocities appears to be reasonably small in most cases. Most results appear to increase or decrease by approximately 15% when initial velocities are altered. The two cases of falls from above standing height, however, appear to show greater scatter in results than the others. Case 4 shows reductions of up to 25%, while Case 3 shows reductions of up to 50%. This could be expected, due to increased space and time over which the initial conditions can take effect.

4.3. Comparing cases to each other

It is not possible from results gained in the present study to conclude that the simulations conducted reveal that certain parameters are correlated with specific brain injuries. This is due to a number of factors including the small number of cases (four), the variety of injury types (almost all types of focal brain injury) and the different head impact locations (two occipital, two temporo-parietal). Also of significance is that all of the cases presented here represent different populations (an elderly lady, a young boy, an adult male and a young lady). If it were possible to concentrate on one population, or more usefully, one lesion type, it may be possible to develop more useful suggestions regarding injury mechanisms. However, the numbers of patients being admitted to the neurosurgery department for simple falls resulting in well defined focal head injury are not abundant. Future work in this area may potentially broaden the accident selection criteria to include other types of accidents (bicycles, sporting accidents, etc.) resulting in similar lesions, or to include other falling accidents which result in diffuse injuries, fatalities and indeed those resulting in mild/absent injuries.

4.4. Comparing results to tolerance limits from the literature

While it may not be possible to draw conclusions regarding injury mechanisms from comparisons of the cases presented here, it is useful to compare the results obtained with tolerance curves proposed by other authors for specific lesion types, as mentioned in the introduction.

When results from the simulations presented here are compared to the upper and lower limits for contusion, acute subdural haematoma and subarachnoid haematoma from Auer *et al.* [21],

and skull fracture tolerance limits from Yoganandan *et al.* [16] and Allsop *et al.* [17], results for the simulations using the default head contact characteristics tend to fall above the upper tolerance curves, or between the lower and upper tolerance curves. Results for almost all of the simulations using the alternative head contact characteristics tend to fall between the two. Thus, in the default head contact cases, simulation results predict a very high likelihood of almost all of the aforementioned lesions occurring in all cases. In the alternative head contact simulations, on the other hand, results do not predict an extremely high probability of occurrence, but do fall into the range where such lesions have been seen to occur.

Thus it appears that the simulations using the alternative head contact characteristics are more realistic than those using the default head contact characteristics with regard to the various tolerance curves and the observed injuries presented here. However, while results here are certainly in line with the tolerance limits suggested by these authors, the lack of case data with values falling below the lower tolerance limits or above the upper tolerance limits makes it impossible to use these results to provide clinical validation for these tolerance limits.

5. Conclusions

The most important factor to be noted from the results is the difference in output obtained when using different head contact characteristics. It appears that the multibody modelling method can supply useful information, *provided* the model is based on the most accurate human response data possible. The default force-penetration response in MADYMO is based on experiments using the EEVC headform and this can overpredict the force associated with a given deformation, when compared against cadaveric experiments on skull bone. In principle, it would be important to have knowledge of a population of head contact characteristics (age, gender, impact location, and strain rate) if skull and brain injuries were to be predicted with precise accuracy. In practice, however, it is not necessary to have patient-specific skull constitutive characteristics although it would be possible to estimate this information from an individual's head CT and bone mineral density data.

Although the results obtained from the present study do not conclusively establish mechanisms of head injury or tolerance to head injury, they do nonetheless confirm that multibody modelling of real life head injury accidents can usefully provide information typically obtained by animal or cadaveric experimental methods, both of which have drawbacks when generalising results to living humans. Results obtained could also provide useful input data for more complex models of the human head, such as a finite element (FE) model [26] [27] [28]. This type of model is able to calculate full-field stresses, strains and strain rates in the brain resulting from impact, and would thus theoretically be able to predict more accurately the location and severity of brain injury resulting from impact.

Although the modelling of *real life* cases of head injury does involve a degree of uncertainty, (i.e., such accidents are not produced in an experimentally controlled environment), the data obtained is a direct contribution to the body of knowledge regarding head injury biomechanics. Further information obtained from such real life head injury cases will serve to establish accurate tolerance limits and injury mechanisms of various lesion types in humans.

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Table 1. Case summaries, providing age, sex, description of accident, and head injury sustained.

Case	Sex	Age	Height (cm)	Weight (kg)	Brief description of accident	Head Injury
1	F	76	160	60	Standing on doorstep (13 cm), lost balance and fell directly backwards. Incurred occipital impact of head against concrete wall 136 cm away from step.	Small left frontal lobe contusion. Large right temporal parenchymal haematoma.
2	M	11	152	37	Fell directly backwards in heat-related fainting episode. Incurred occipital impact of head on concrete pavement.	Right frontal lobe contusion. Traumatic subarachnoid haemorrhage.
3	M	37	178	80	Fell backwards and twisted to left while balancing on gate (138 cm), pulling a rope which failed. Incurred left lateral impact of head on tarmac.	Left temporo-parietal linear skull fracture. Left temporo-parietal extradural haematoma.
4	F	24	169	55	Standing on chair (44 cm), twisted sharply to right in response to noise and fell forwards and to right. Incurred right lateral impact of head on ceramic tiled floor.	Right frontal linear skull fracture. Right frontal extradural haematoma. Left posterior temporal basal contusion.

Table 2. Peak results of simulations run using default head contact characteristics.

	Linear Velocity (m/s)	Angular Velocity (rad/s)	Linear Acceleration (g)	Angular Acceleration (krad/s ²)	Force (kN)	HIC
Case 1	4.87 - 5.31	32.56 - 35.84	311 – 350	44.6 – 50.2	9.9 - 11.1	1153 - 1479
Case 2	5.41 – 5.80	41.04 – 49.10	638 – 714	44.8 – 64.7	22.3 – 24.7	5834 – 7671
Case 3	4.44 – 7.11	28.04 - 40.56	463 – 1015	26.2 – 54.3	22.5 – 47.4	3316 – 15027
Case 4	3.97 – 5.43	27.42 – 35.11	443 – 612	42.6 – 61.8	14.8 – 27.7	2514 - 5933

Table 3. Peak Results for simulations run using alternative head contact characteristics.

	Linear Velocity (m/s)	Angular Velocity (rad/s)	Linear Acceleration (g)	Angular Acceleration (krad/s ²)	Force (kN)	HIC
Case 1	4.87 - 5.31	32.49 - 35.81	243 – 293	34.7 – 41.4	7.8 – 9.3	821 - 1057
Case 2	5.41 – 5.80	41.01 – 49.10	360 – 387	41.1 – 43.5	13.2 – 14.2	3355 - 3978
Case 3	4.44 – 7.11	27.04 – 37.75	268 – 435	17.6 – 42.3	13.6 – 21.6	1739 - 5580
Case 4	3.97 – 5.43	27.42 – 35.09	271 – 334	27.7 – 32.9	10.4 – 14.5	1560 - 2860

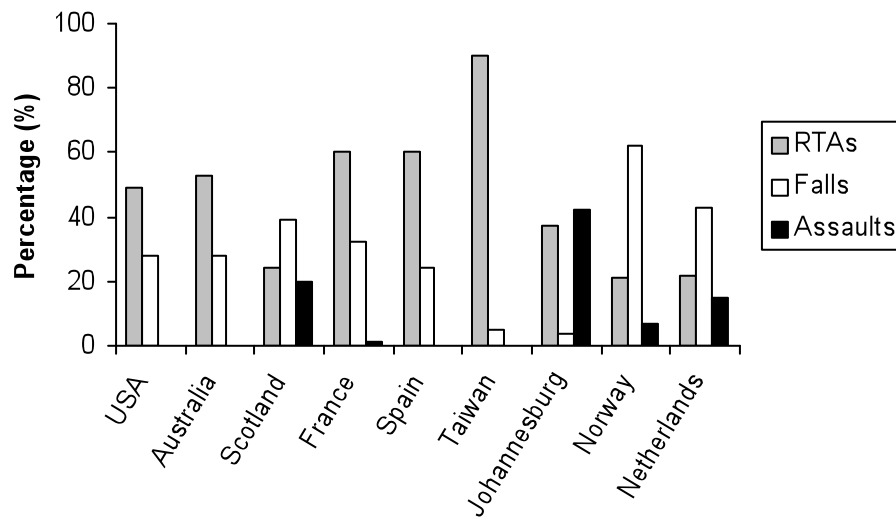


Fig. 1. Distribution of causes of head injury worldwide [1,3-5]. Where no column is represented, there is no information for that statistic.

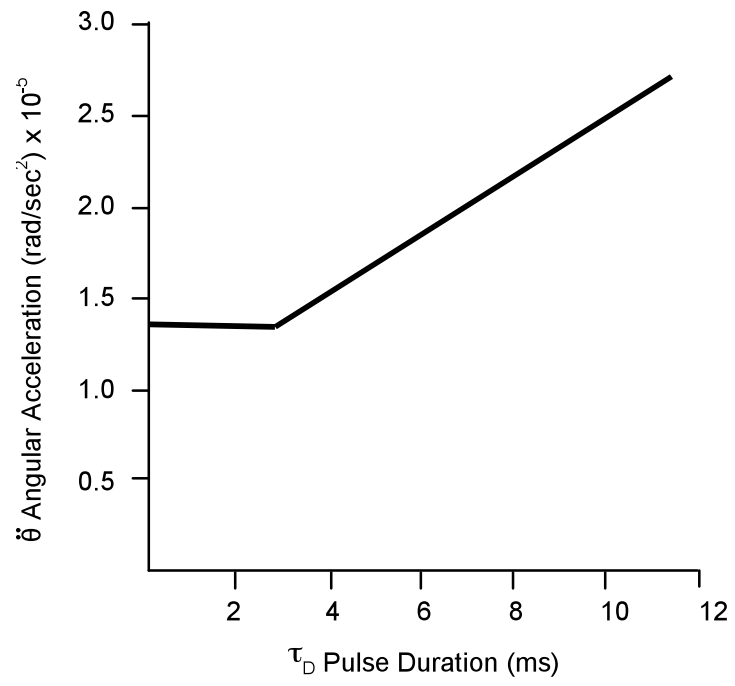


Fig. 2. Gennarelli and Thibault's [11] tolerance curve for ASDH in rhesus monkeys.

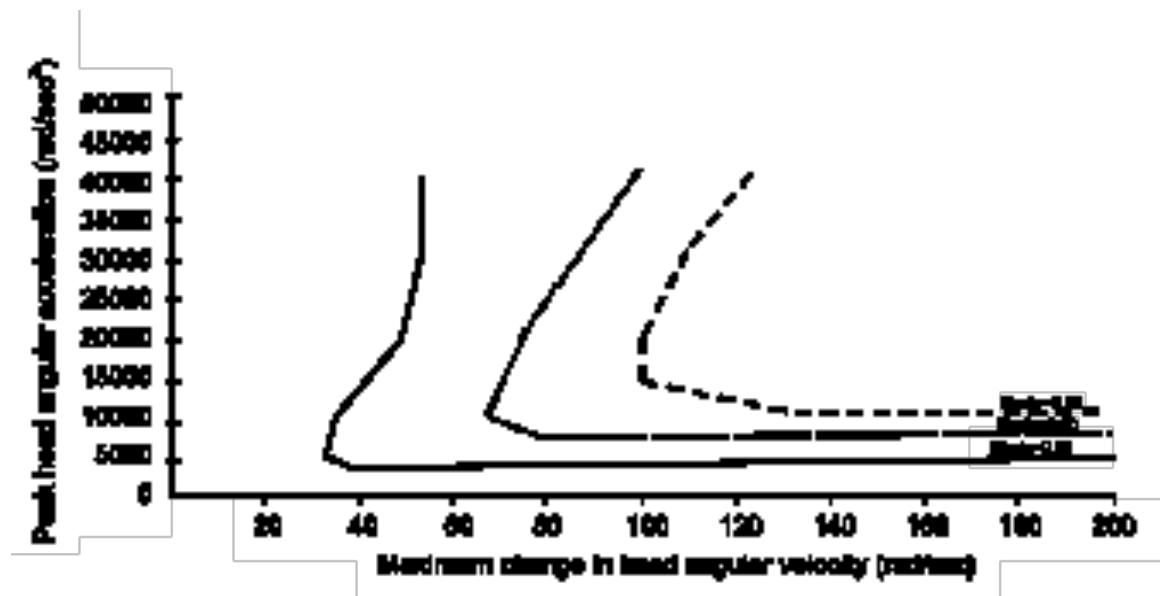


Fig. 3. Tolerance curves for DAI and cerebral concussion (reproduced from Ryan and Vilenius [15]).

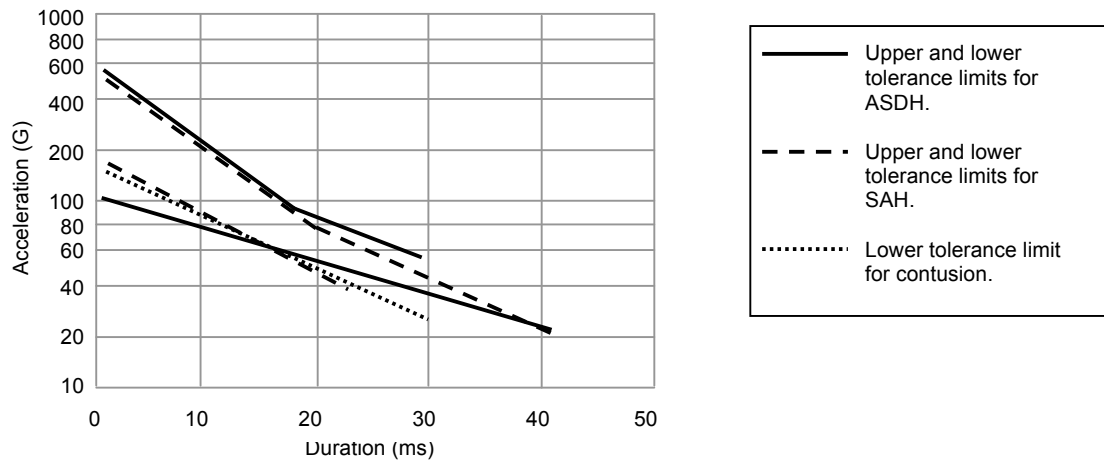


Fig. 4. Upper and lower tolerance curves for ASDH, subarachnoid haematoma and contusion [21].

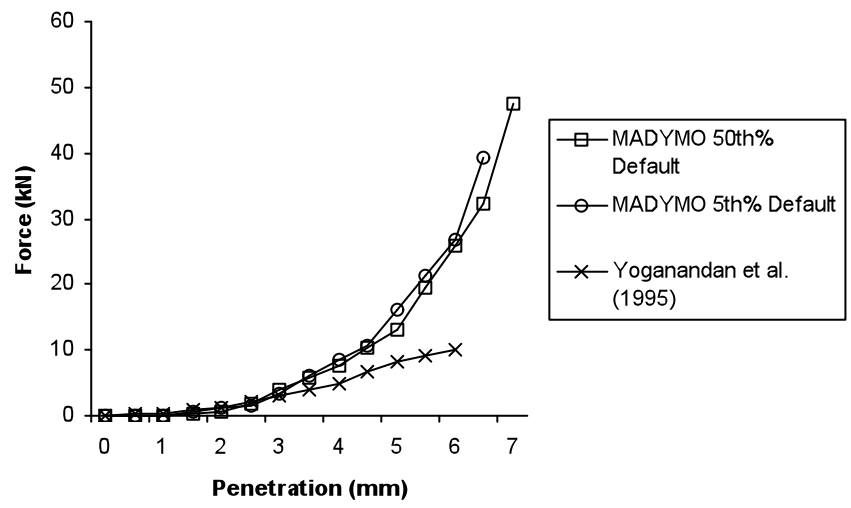


Fig. 5. Head contact force-penetration curves.

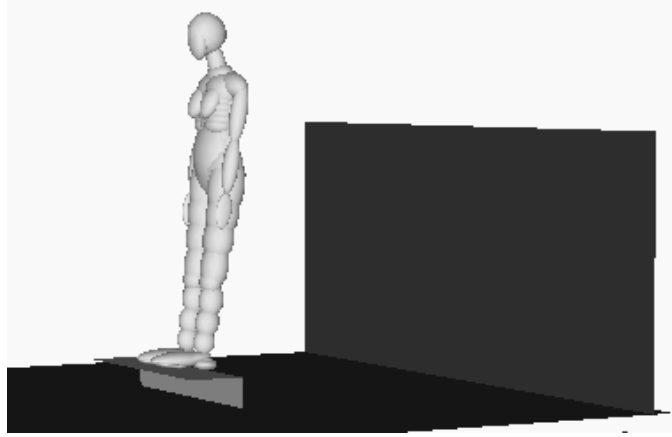


Fig. 6a. Initial position of Case 1 described in Table 1.

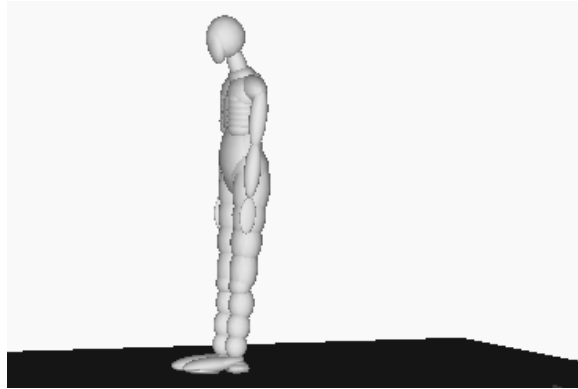


Fig. 6b. Initial position of Case 2 described in Table 1.

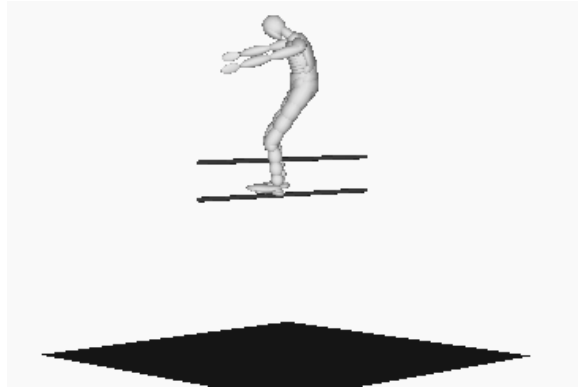


Fig. 6c. Initial position of Case 3 described in Table 1.

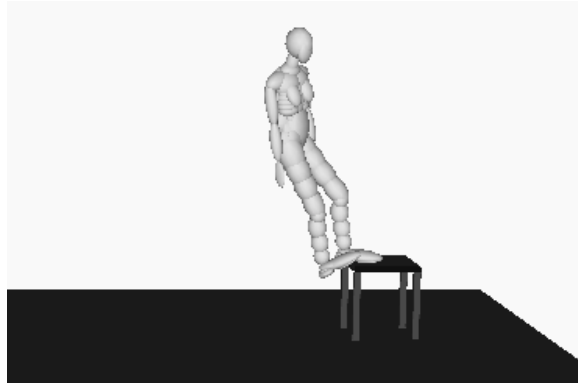


Fig. 6d. Initial position of Case 4 described in Table 1.

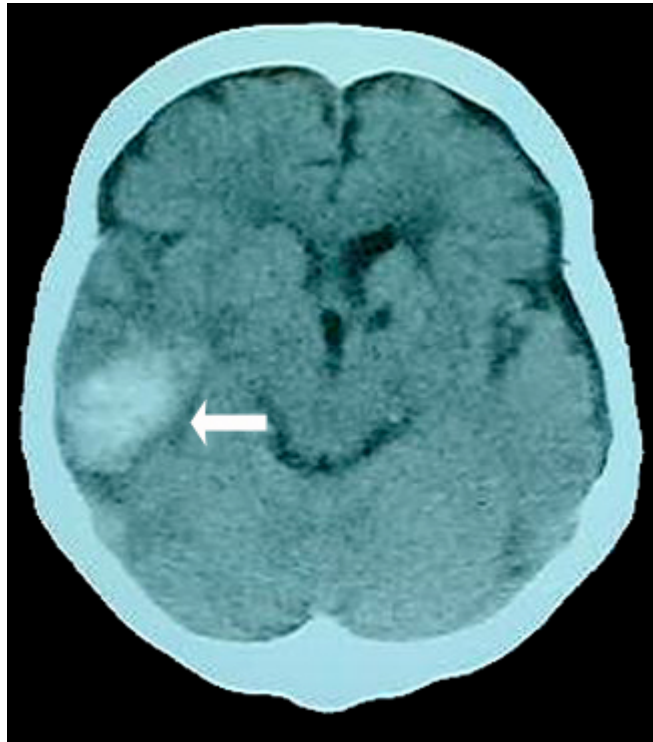


Fig. 7a. 6 hour CT scan of Case 1 described in Table 1. Arrow indicates large right temporal parenchymal haematoma.

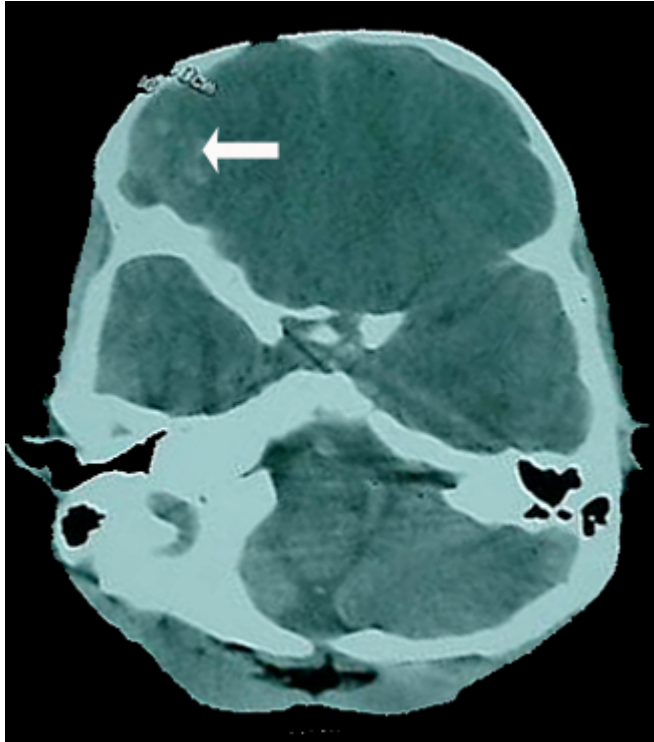


Fig. 7b. 6 hour CT scan of Case 2 described in Table 1. Arrow indicates right frontal lobe contusion.

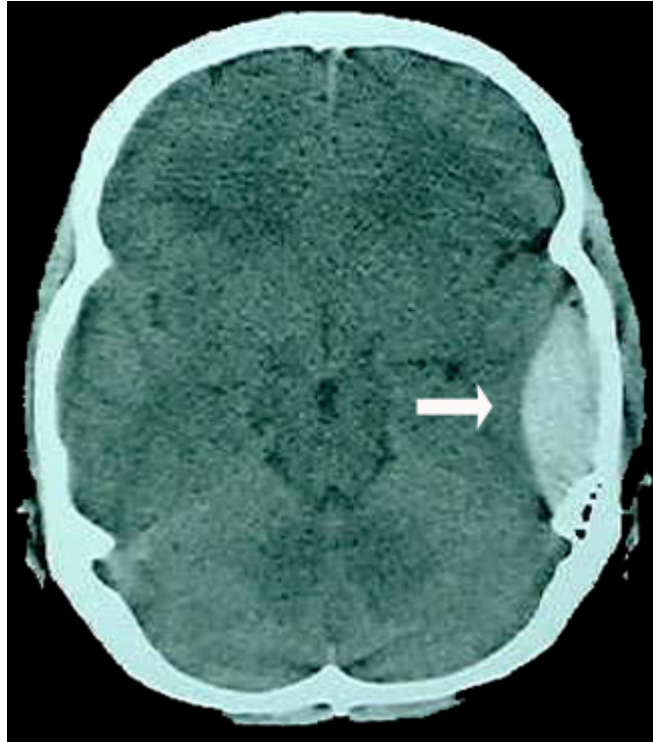


Fig. 7c. 6 hour CT scan of Case 3 described in Table 1. Arrow indicates left temporo-parietal extradural haematoma.

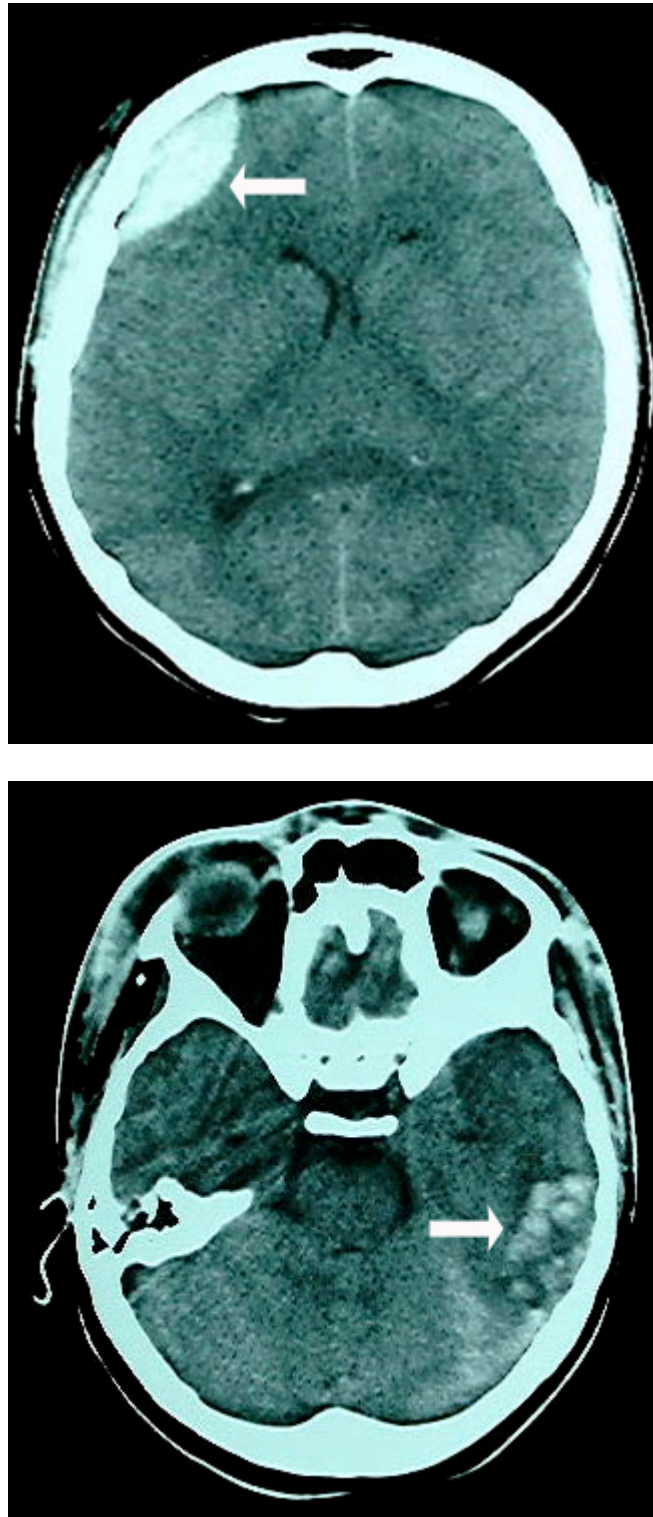


Fig. 7d. 6 hour CT scans of Case 4 described in Table 1. Arrow in top scan indicates right frontal haematoma whilst arrow in bottom scan indicates left posterior temporal basal contusion.

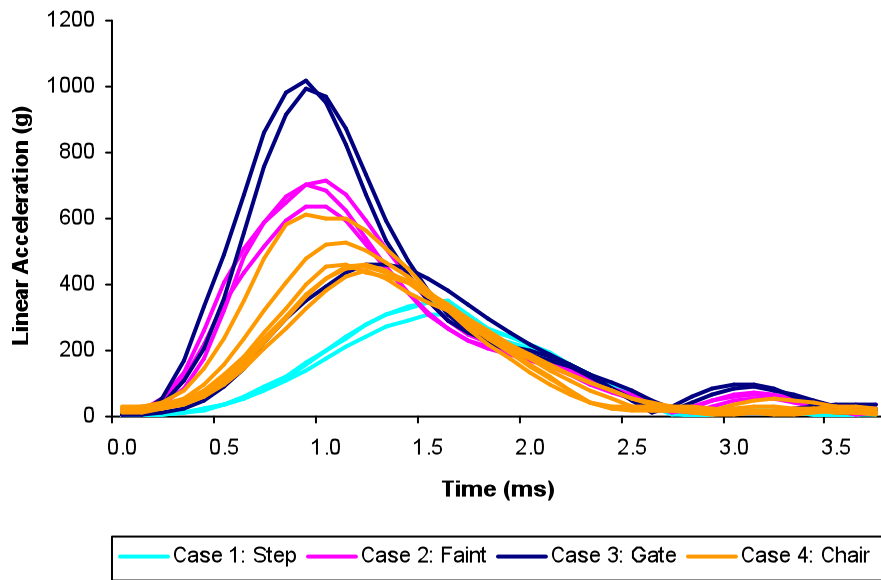


Fig. 8 (a). Linear acceleration for all simulations using default head contact characteristics.

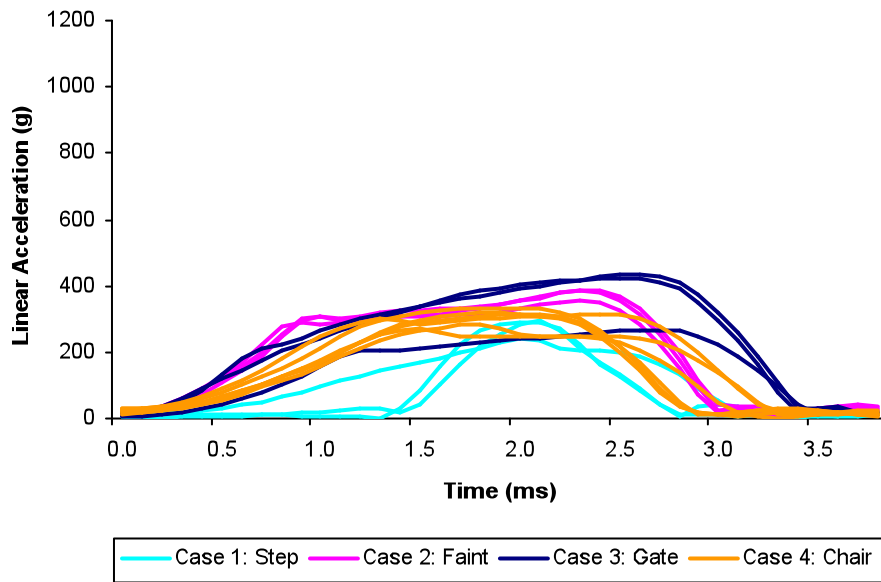


Fig. 8 (b). Linear acceleration for all simulations using alternative head contact characteristics.

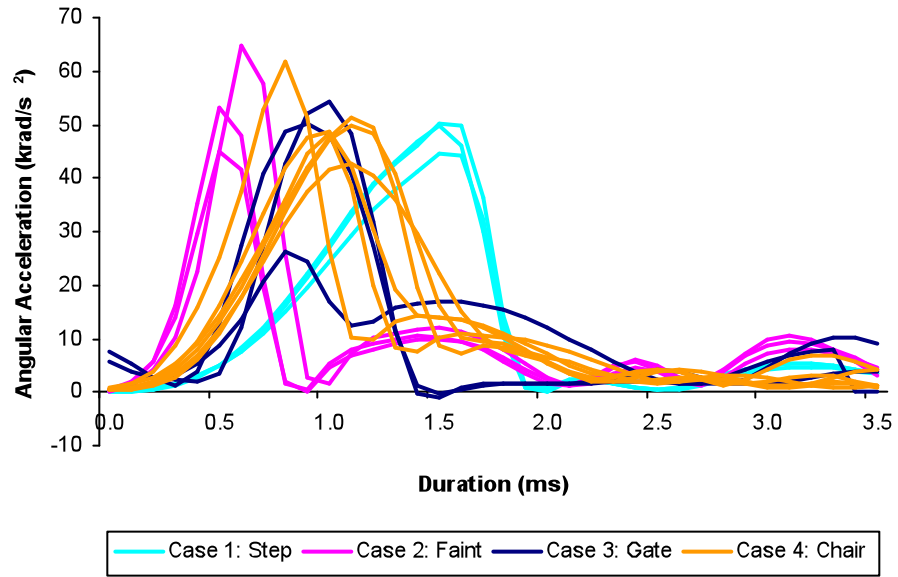


Fig. 9 (a). Angular acceleration for all simulations using default head contact characteristics.

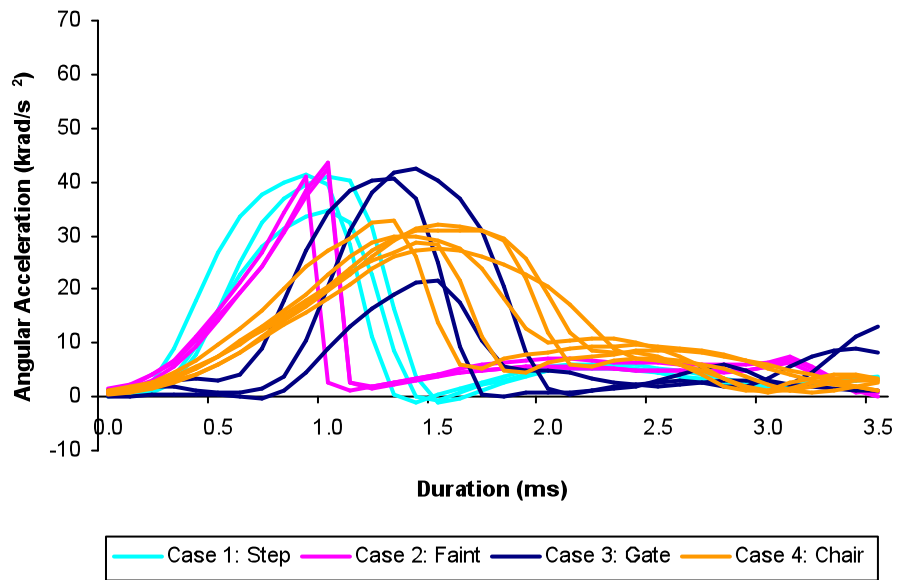


Fig. 9 (b). Angular acceleration for all simulations using alternative head contact characteristics.

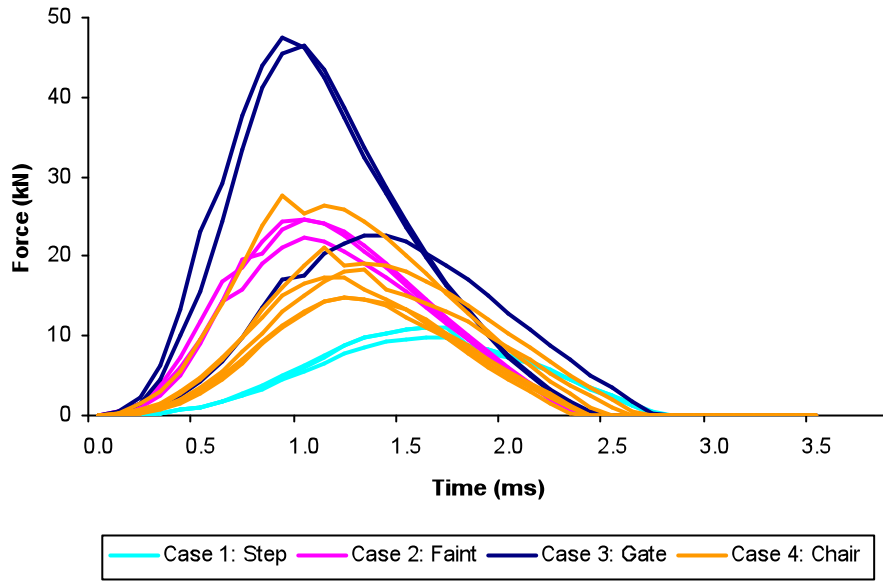


Fig. 10 (a). Contact force on head for all simulations using default head contact characteristics.

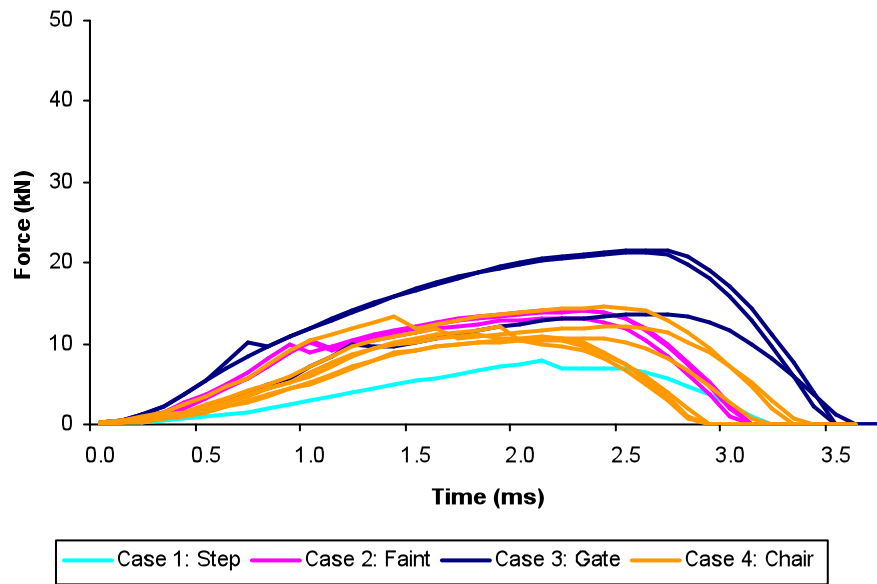


Fig. 10 (b). Contact force on head for all simulations using alternative head contact characteristics.