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ORIGINAL ARTICLE

**Jean-Sébastien Raul · Daniel Baumgartner ·
Rémy Willinger · Bertrand Ludes**

Finite element modelling of human head injuries caused by a fall

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Abstract Finite element models (FEMs) can be used as prediction tools for human head injuries caused by falls. The purpose of this paper is to demonstrate the relevance of using human head FEM to assess the possible mechanism for the origin of head injuries. The FEM of the human head used in this study was developed in the late 1990s at the University Louis Pasteur of Strasbourg (ULP) and has been validated for human head impacts for simulating human head injuries caused by car accidents. Its use in legal medicine appears to be very useful for comparing different injury mechanisms. We present the simulation obtained for two witnessed falls of the same individual, and compare our results to tolerance limits of the main human head injuries. We show that this tool can be used to discuss the possible mechanism of injury encountered for the observed lesions in a forensic case. It can also help to distinguish between possible and impossible human head injury mechanisms.

Keywords Finite element models · Forensic science · Head injury · Reconstruction

J.-S. Raul · B. Ludes
Institut de Médecine Légale,
11 rue Humann,
67085 Strasbourg, France

D. Baumgartner · R. Willinger
Institut de Mécanique des Fluides et des Solides,
UMR 7507 ULP CNRS,
2 rue Boussingault,
67085 Strasbourg, France

J.-S. Raul (✉)
Institut de Médecine Légale,
11 rue Humann,
67085 Strasbourg, France
e-mail: Jean-Sebastien.Raul@iml-ulp.u-strasbg.fr
Tel.: +33-39-0243361
Fax: +33-39-0243362

Introduction

Until recently the legal system relied on the testimony of medical experts to determine whether the force imparted to the human head in a given scenario was consistent with the resulting human head injury. Since the 1960s predictive human head impact indices have been developed to aid the investigation of causation of human head injury. Finite element models (FEMs) can serve as interesting tools for the forensic scientists when various human head injury mechanisms need to be evaluated (e.g. adult and child head injuries, ballistic human head injuries). Human head FEMs are mainly used in car crash evaluations and are not commonly used in forensic science. Recent technological progress has resulted in the creation of simpler tools which can be used in forensic cases. To illustrate this purpose and to show some of the possibilities of human head FEMs, we performed the reconstruction of two well-documented falls and compared the results to clinical data.

Case report

A 63-year-old male alcoholic was slightly pushed by a young adult after annoying his girlfriend during a party in France. Being drunk, he took two steps backward, stopped, and fell backwards like a stick, thus sustaining a parieto-occipital head impact on the wooden floor. This fall was witnessed by ten people who all gave the same version of the fall. The victim immediately lost consciousness. The nearest hospital was called, and three firemen arrived 10 min later and placed the victim on a cart. As the cart was being lifted to a height of about 20–30 cm, the victim fell from the cart on his right side, sustaining a parietal head impact.

After arriving in the emergency room, the victim was taken to a critical care unit. A CT scan was performed; results showed right and left frontal, temporal and occipital contusions, an occipital subdural haematoma, diffuse arachnoid haemorrhage, and a parieto-occipital skull fracture. In spite of 1 month of intensive care, the victim finally died and an autopsy was performed. The skull fracture was still visible during autopsy and the basal part of the temporal and frontal lobes of the brain showed necrotic evolution of the contusions seen on the CT scan.

Materials and methods

ULP finite element of the human head

The FEM of the human head used in this study (called ULP FEM of the human head) was developed by Willinger et al. at the University Louis Pasteur of Strasbourg (ULP) [1]. It is assumed to be homogenous and isotropic. It is continuously meshed in 11,939 nodes and 13,208 elements, which are subdivided in 10,395 brick elements and 2,813 shell elements. Its geometrical characteristics and the mechanical behaviour of its anatomical components are as follows:

A viscous elastic brain and cerebellum (Boltzman model) meshed in 5,508 brick elements. Density=1,040 kg m^{-3}, bulk modulus=1,125 MPa, short time shear modulus=0.049 MPa, long time shear modulus=0.0167 MPa, decay constant=145 s^{-1}. The mechanical behaviour relies on experimental data from Schuck and Advani [2].

An elastic subarachnoid space (linear model) meshed in 2,591 brick elements. Density=1,040 kg m^{-3}, Young modulus=0.012 MPa, Poisson ratio=0.49. The mechanical behaviour relies on experimental data from Willinger et al. [3].

An elastic falx of the brain (linear model) and an elastic tentorium of the cerebellum (linear model) meshed in 471 shell elements. Density=1,140 kg m^{-3}, Young modulus=31.5 MPa, Poisson ration=0.23. The mechanical behaviour relies on experimental data from Zhou et al. [4].

An elastic plastic brittle skull (Tsai Wu model) and an elastic plastic brittle face (Tsai Wu model) meshed in 1,813, three-layered composite shell elements (cortical bone–trabecular bone–cortical bone). Cortical bone: density=1,800 kg m^{-3}, Young modulus=15 GPa, Poisson ration=0.21, ultimate tensile stress=90 MPa, ultimate compression stress=145 MPa. The mechanical behaviour relies on experimental data from Wood [5]. Trabecular bone: density=1,500 kg m^{-3}, Young modulus=4.5 GPa, Poisson ration=0, ultimate tensile stress=35 MPa, ultimate compression stress=28 MPa. The mechanical behaviour relies on experimental data from Melvin and Evans [6].

An elastic scalp (linear model) meshed in 2,296 brick elements. Density=1,200 kg m^{-3}, Young modulus=16.7 MPa, Poisson ratio=0.42. The mechanical behaviour relies on experimental data from Zhou et al. [4].

The ULP FEM of the human head is validated in terms of brain pressure and brain acceleration from the experimental data of Nahum et al. [7] and Trosseille et al. [8], and in terms of bone rupture (i.e. skull fracture) from the experimental data of Yoganandan et al. [9]. The mechanical behaviour of the modelled anatomical components of the ULP FEM of the human head is summarised in Table 1.

The FEM of the human head was developed by using the HYPERMESH code for geometry and meshing, and the RADIOSS CRASH code for numerical simulation. This code allows dynamic simulations to be achieved in which the geometrical deformation is taken into account at each time step. Elements used in the ULP FEM of the human head are as follows:

Shell elements with three nodes (tria elements) or four nodes (quad elements).

Brick elements with six nodes (hexa elements) or eight nodes (tetra elements).

Table 1 Mechanical behaviour of the ULP finite element model of the human head (modelled anatomical components)

Anatomical component	Mechanical factors								
	Density (kg/m^3)	Young modulus (MPa)	Poisson ratio	Ultimate compression stress (MPa)	Ultimate tensile stress (MPa)	Bulk modulus (MPa)	Short time shear modulus (MPa)	Long time shear modulus (MPa)	Decay constant (s^{-1})
Brain and cerebellum	1,040	–	–	–	–	1,225	0.049	0.0167	145
Subarachnoid space	1,040	0.012	0.49	–	–	–	–	–	–
Falx of the brain	1,140	31.5	0.23	–	–	–	–	–	–
Tentorium of the cerebellum	1,140	31.5	0.23	–	–	–	–	–	–
Cortical bone	1,800	1,500	0.21	145	90	–	–	–	–
Trabecular bone	1,500	4,500	0.0	28	35	–	–	–	–
Scalp	1,200	16.7	0.42	–	–	–	–	–	–

These elements have one integration point, and a dynamic Lagrange formulation was used since each modelled part is assumed to have a homogenous and isotropic solid mechanical behaviour. The meshing was formed in a regular way in terms of element dimension, angle and warpage. The average dimension of the ridge of an element is 5 mm. The meshing is continuous between the different anatomical components that were modelled, therefore no interface was used in the human head FEM.

A numerical reconstruction of the two falls was performed using the ULP FEM of the human head to assess the consequences of each fall. The mechanic properties of the wooden floor were taken into account for the simulation as follows. A flat surface was regularly meshed in shell elements with a thickness of 10 mm. A density of 0.9 kg m^{-3}, a Young modulus of 11 GPa and a Poisson ratio of 0.49 were used for the linear elastic behaviour of the wooden floor. The wooden floor was also assumed to be stuck to a non-deformable surface representing the base floor. The initial velocity for the human head was 6 m s^{-1} for the first fall, and 1.5 m s^{-1} for the second fall. That initial velocity was inferred from an analytical analysis, using general mechanics equations, of both falls of the victim. Both velocities were applied to the FEM of the head of the victim. After those initial conditions, the head of the victim is left free in its 6 degrees of freedom both in translation or in rotation. It is important to underline that the neck does not influence the mechanics of the human head in the first 30 ms after the impact. Both impacts under study lasted no more than 15 ms. Therefore the human head can be considered as free of motion just after each impact on the floor. The locations of the human head impact were also taken into account for the reconstruction. The first impact of the head of the victim was a right occipito-parietal impact, the second was a right parietal impact.

Mechanical field parameters sustained by the different tissues of the head of the victim during the impact were obtained after simulation, such as the brain pressure and the Von Mises stress of the brain (which represents the shearing stress sustained by the brain), the global strain energy of the skull and the number of deleted elements, and the global strain energy of the subarachnoid space.

Human head injury mechanisms and tolerance limits

The human head injury mechanisms and tolerance limits used in the present study were derived from a numerical real-world accidents reconstruction [10], in which 64 accidents of helmeted motorcyclists and American football players, as well as unprotected pedestrians, were reconstructed. The ULP FEM of the human head was submitted to the kinematics, in terms of accelerations, sustained by each accident victim, and measured experimentally using an accident replication with a HYBRID III dummy head. In this study, we established a correlation between the observed injuries and calculated mechanical parameters using the ULP FEM of the human head. The main results of that study are as follows:

- A brain pressure reaching 200 kPa is an indicator for brain contusions, oedema and haematoma.
- A brain Von Mises stress reaching 18 kPa is an indicator for moderate neurological injuries.
- A brain Von Mises stress reaching 38 kPa is an indicator for severe neurological injuries.
- A global strain energy of the subarachnoid space reaching 5.4 J is an indicator for subdural haematoma and subarachnoid bleeding.
- A global strain energy of the skull reaching 2.2 J is an indicator for skull fractures.

These human head tolerance limits are summarised in Table 2.

Results

Both numerical simulations lasted 8 h on a DEC ALPHA server. The first 15 ms of each head impact are simulated. The average time step is of 10^{-3} ms, which is compatible with the propagation phenomenon in the modelled materials. This time step is calculated by the RADIOSS CRASH code and calculated at each new time increment. This time step calculation depends on the size of the smallest elements and on the sound velocity in the different modelled materials.

Table 2 Calculated mechanical parameters in comparison to the tissues tolerance limits

Conditions	Mechanical parameters			
	Brain pressure (kPa)	Brain Von Mises stress (kPa)	Global strain energy of the skull (J)	Global strain energy of the subarachnoid space (J)
Tolerance limit	200	18	2.2	5.4
Indication	Brain haemorrhages	Brain neurological lesions	Bone ruptures	Subarachnoid and subdural haematoma
Fall configuration 1	574	66	27.5	14.7
Fall configuration 2	53	12	0.6	0.5

The results are summarised in Table 2. For the first impact configuration under study, the brain pressure reached values of 574 kPa, predicting brain contusion. The human head model also predicted the location of the bleeding and showed a classical coup and contre-coup injury located in the occipital (for the coup) and fronto-temporal (for the contre-coup) lobes. As seen in Figs. 1 and 2, the distribution of brain pressures correlated with the images obtained from the CT scan performed when the victim was admitted to hospital. For the second impact configuration, the brain pressure remained below the tolerance limit. Therefore, brain contusion should not be observed after this impact.

The first impact configuration predicted a coma with a calculated Von Mises stress of 66 kPa. This immediate loss of consciousness was witnessed by surrounding observers. For the second configuration under study, the brain Von Mises stress remained below the proposed tissue tolerance limit (12 kPa). Therefore, moderate neurological injuries should not be observed in this case.

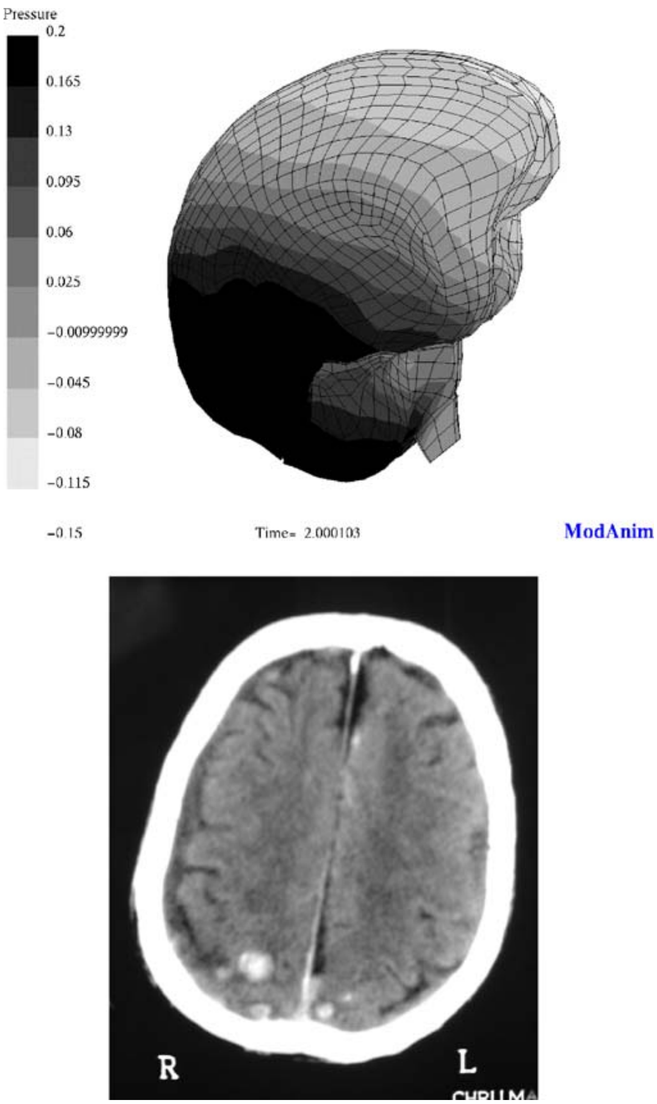

Fig. 1 Results of the brain pressure value and anatomical distribution 2 ms after the beginning of the impact (*upper diagram*) compared to imagery (*lower diagram*). Prediction of coup lesions

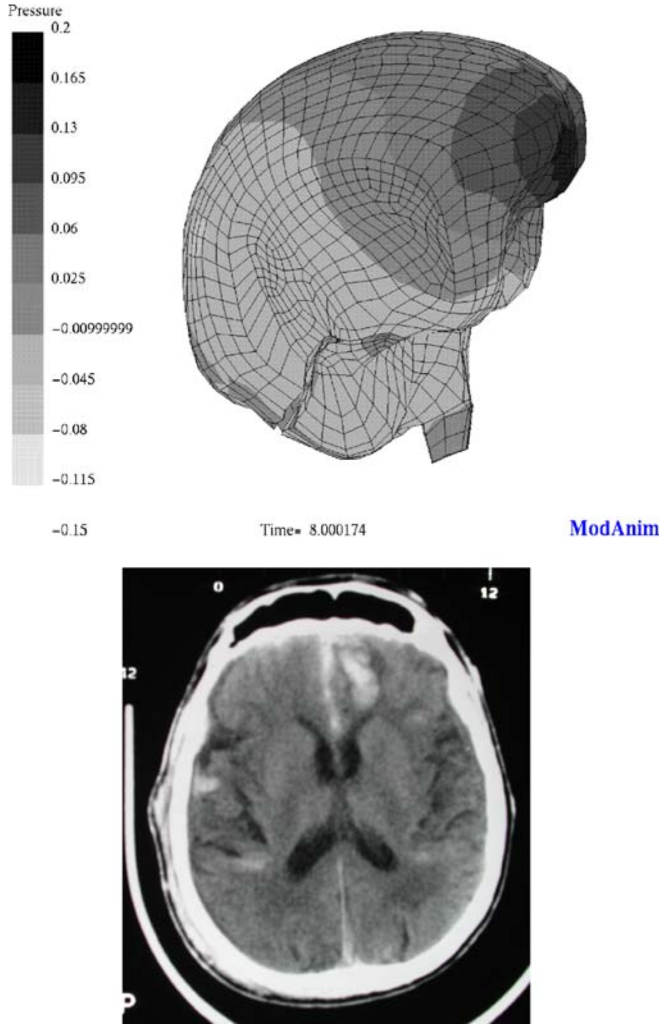

Fig. 2 Results of the brain pressure value and anatomical distribution 8 ms after the beginning of the impact (*upper diagram*) compared to imagery (*lower diagram*). Prediction of contre-coup lesions

For the first impact configuration, the calculated global strain energy of the skull reached a value of 27.5 J and bone fracture was predicted by the model through the destruction of 40 of the three-layered composite shell elements of the skull, as shown in Fig. 3. The anatomical distribution of the deleted elements is close to the fracture seen at autopsy, yet it is not very accurate. The reason is that the skull of our model has the same thickness everywhere. A new skull is being developed in our institution which takes into account the different thickness of the different areas of the skull for a better correlation between the predicted and real-life skull fractures. For the second impact configuration, the calculated global strain energy of the skull was of 0.6 J and no three-layered composite shell elements of the skull were destroyed.

For the first impact configuration, the global strain energy of the subarachnoid space reached values of up to

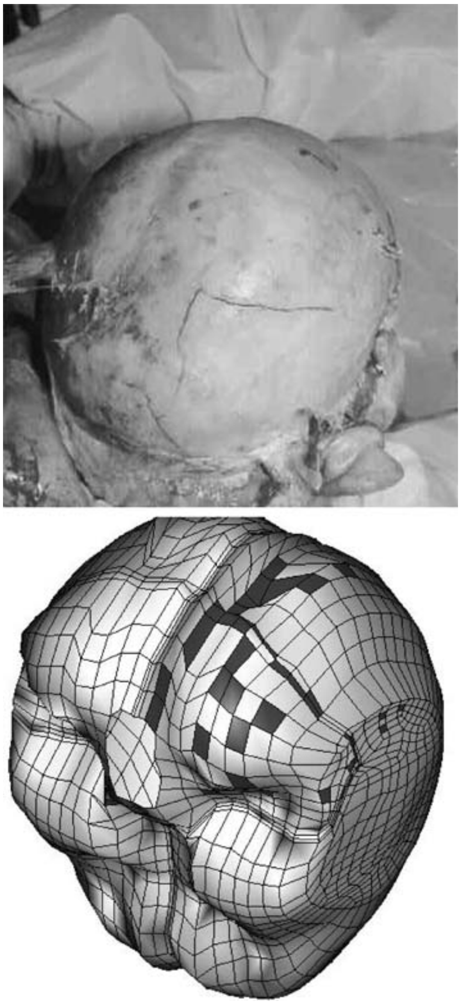

Fig. 3 Comparison of the skull fracture location predicted by reconstruction (*lower diagram*) and the skull fracture observed at autopsy (*upper diagram*). The deleted elements obtained after reconstruction are figured in *black colour*

14.7 J, 20 times higher than for the second impact configuration. Subdural haematoma and subarachnoid bleeding are therefore very likely to occur in the first impact configuration.

Discussion

In the last 40 years, biomechanical studies have been conducted in the field of forensic research. Many of these studies were aimed at (1) establishing whether an infant's head injury was caused by accident or abuse, and if the results indicate abuse, (2) determining the possible mechanisms leading to certain injuries such as subdural haematomas [11, 12]. These studies have been recently criticised [13, 14]. Other works, such as multi-body dynamic reconstructions of adult head injury accidents and biomechanical studies of falls, have recently been published [15–18].

FEMs are considered to be the best and the most powerful tools to investigate the dynamic response of the human head under impact conditions. Nevertheless, no work has yet been published using FEMs for the evaluation of human head injuries in forensic cases.

Car safety standards rely upon criteria for human tolerance, which are based on biomedical research performed more than 30 years ago. Measures designed to improve human head protection are typically evaluated against a measurement of the head injury criterion (HIC). The predictive capacity of this criterion has been widely criticised because of its limited ability to predict brain injury or other specific injuries. It has been suggested that specific deformation or stress of the skull and brain tissue and a measure of the relative motion of the brain and skull would be much better to assess human head protection.

To date, more than ten different three-dimensional human head FEMs have been described, but so far only Ruan et al.'s [19] and Zhou et al.'s [4] models were validated and used for accident reconstruction to investigate brain injury tolerance limits. Most of the essential anatomical components of the human head were incorporated in their models. These models were validated against the brain pressure data from experimental impact tests onto the front of the head of cadavers and against brain skull relative motion using data obtained from high-speed X-ray experiments.

An improved version of the Wayne State University (WSU) model has been published by Zhang et al. [20]. A refined brain meshing was proposed (meshed in 314,500 elements) and the validation procedure showed realistic results for linear and rotational accelerations up to 200 g and 1,200 rad s^{-2}, respectively. Bandak and Chan [21] also published a finely meshed human head FEM and proposed an in-depth experimental and numerical analysis of the subarachnoid cerebral spinal fluid (CSF) layer influence on the brain response.

Moreover, Zhou et al. [4] simulated a fully documented road accident with their model and the shear stresses predicted agreed approximately with the location of axonal injury described by the medical report. More recently, Newman et al. [22] presented a detailed methodology for the assessment of mild traumatic brain injury based on the reconstruction of accidents of the American National Football League using the Zhou et al. [4] human head FEM.

The WSU brain injury model was also used to correlate brain Von Mises shearing stresses with angular head acceleration and brain pressure with the head linear acceleration. Correlation coefficients of 0.86 and 0.82, respectively, were found in this study by Yang and King [23]. In 2001, Bandak et al. [24] presented the first version of a human head injury assessment tool based on a very simplified human head FEM called Simulated Injury Monitor (SIMON).

Currently, real-world accident analysis is used in an attempt to correlate a known human head injury parameter with the Abbreviate Injury Scale (AIS) value sustained. An attempt by Chinn et al. [25] to correlate initial human head impact velocity, maximum linear and rotational acceleration, HIC value and General Acceleration Model for Brain Injury Threshold (GAMBIT) value (that takes into account not only linear acceleration but also rotational acceleration of the human head) versus AIS gave a correlation coefficient range of 0.3–0.6, which is not satisfactory. The main reason for the poor correlation between a given parameter and AIS was that the same AIS levels can be reached from very different injury mechanisms. A much better approach is to take into account the likelihood of the human head injury mechanism.

When the type of lesion, rather than the AIS, is used for comparison, the brain Von Mises stress is a good indicator for brain neurological lesions (whether they are moderate or severe). Moreover, this mechanical parameter allows these lesions to be distinguished into two categories: moderate or severe. Global strain energy in the CSF layer and in the skull structure is a reasonable indicator for subdural haematoma and skull fracture, respectively.

The tolerance limits used for the ULP FEM of the human head have been detailed in previous works [10, 26]. Brain pressure has been correlated with brain haemorrhages resulting in brain contusions, oedema and haematoma when values exceeded 200 kPa. The tolerance limits predicting neurological injuries resulting in unconsciousness, contusion and coma are correlated to the calculated brain Von Mises stress. Tolerance limits for neurological moderate or severe injuries with a 50% risk are established for brain Von Mises stress of 18 and 38 kPa, respectively. These values can be compared to previously reported data such as 11 kPa by Zhou et al. [4], 15 kPa proposed by Kang et al. [27], or 27 kPa suggested by Anderson [28]. The global strain energy of the subarachnoid space has been correlated with haemorrhages resulting in subdural and subarachnoid haemorrhages when values exceeded 5.5 J with a 50% risk of occurrence. In other studies, this injury was evaluated by the parasagittal elongation of bridging veins or their elongation rate computed with the FEM, as suggested by Bandak et al. [24]. Tolerance limits for skull fracture have been reported from experimental data in terms of impact force (4–14 kN) by Yoganandan et al. [9] as well as in terms of strain energy (around 2 J) by Gurdjian and Webster. [29]. For the ULP FEM of the human head used in this study, a tolerance limit for skull fracture was established numerically at 2.2 J with a 50% risk. This energy represents the global strain energy of the subarachnoid space.

We studied two impacts that happened to the same individual, with medical examination being performed after the second impact which could have caused a problem in separating the effects of each impact. This would certainly be true if the energies of both impacts and the location of the impacts were close. In the case under study, both velocities at the time of impact were four times different (6 m s^{-1} for the first impact and 1.5 m s^{-1} for the second impact), and the location of the first (occipital) and second impacts were also different (a side impact). The location of the skull fracture seen on the victim was more likely explained by the first fall. Also, there was a witnessed immediate loss of consciousness following the first fall. The results of the simulation of the first fall predicted all of these findings. The results of the simulation of the second fall showed brain pressures ten times lower than the first fall and four times lower than the tolerance limits, and global strain energies of the subarachnoid space was 30 times lower than in the first fall and ten times lower than the tolerance limits. The first fall was therefore very much more likely to create the contusions seen on the CT scan. Moreover, the simulation of the first fall showed an anatomical distribution of predicted brain contusions comparable to the contusions seen on the CT scan. The point was to compare the two falls, and it can be concluded that the second fall alone was very unlikely to have created any injury seen on the first CT scan of the victim. However, it is of course difficult to discuss the consequences of the second fall on an injured human head, even if the predicted levels of the mechanical field parameters are very much lower than the tolerance limits values. Conclusions can, of course, only be drawn for this particular case.

The ULP FEM of the human head can also be used when completing clinical data to compare the biomechanical consequences of a fall to blows when a victim is found unconscious or dead, and these two scenarios are being discussed.

The ULP FEM of the human head is a head model. Therefore only impacts to the human head can be discussed with minor levels of uncertainty. The possible effects of impacts to the trunk or to the arms and legs preceding the impact to the human head can be taken into account, but may raise many questions concerning the energy of the impact to the human head. Therefore, the biomechanical study can be completed by the use of multi-body dynamics.

A biomechanical approach can be very helpful to investigate forensic cases and there is a need for collaboration between forensic sciences and biomechanics to objectively and scientifically evaluate adult and infant head injuries using well-documented cases.

Conclusion

The ULP FEM of the human head was developed in the late 1990s and has been validated by experimental impacts on cadavers. The present study shows the relevance of numerical methods, and especially FEM, for reconstructing human head injuries. For the case studied, two falls were compared and their consequences were shown to be very different. Moreover, the injuries predicted by the human head ULP FEM were very similar to the injuries sustained by the victim. Numerical tools have to be accurately defined in terms of geometry and mechanical behaviour. FEMs may then become an invaluable tool in the future for injury prediction and may therefore contribute in the objective and scientific evaluation of adult and infant head injuries in forensic cases.

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