Study of cerebrospinal injuries by force transmission secondary to mandibular impacts using a finite element model

Lucile Tuchtan, Yves Godio-Raboutet, Clémence Delteil, Georges Léonetti, Marie-Dominique Piercecchi Marti, Lionel Thollon

Study of cerebrospinal injuries by force transmission secondary to mandibular impacts using a finite element model

Lucile Tuchtan [a,b,*], Yves Godio-Raboutet [c,d], Clémence Delteil [a,b], Georges Léonetti [a,b], Marie-Dominique Piercecchi Marti [a,b], Lionel Thollon [c,d]

a *Forensic department, APHM, Hôpital de la Timone, 13385 Marseille, France*

b *Aix Marseille Univ, CNRS, EFS, ADES, Marseille, France*

c *Aix Marseille Univ, IFSTTAR, LBA, Marseille, France*

d *iLab-Spine (International Laboratory – Spine Imaging and Biomechanics)*

* Corresponding author. Tel.: +33 491482245; fax: +33 491923331

E-mail address: lucile.tuchtan@ap-hm.fr (Lucile Tuchtan)

Abstract

Brain and cervical injuries are often described after major facial impacts but rarely after low-intensity mandibular impacts. Force transmission to the brain and spinal cord from a mandibular impact such as a punch was evaluated by the creation and validation of a complete finite element model of the head and neck. Anteroposterior uppercut impacts on the jaw were associated with considerable extension and strong stresses at the junction of the brainstem and spinal cord. Hook punch impacts transmitted forces directly to the brainstem and the spinal cord without extension of the spinal cord. Deaths after this type of blow with no observed histological lesions may be related to excessive stressing of the brainstem, through which pass the sensory-motor pathways and the vagus nerve and which is the regulatory center of the major vegetative functions. Biological parameters are different in each individual, and by using digital modeling they can be modulated at will (jaw shape, dentition…) for a realistic approach to forensic applications.

Keywords: Finite element model, Cerebrospinal injury, Force transmission, Brainstem, Mandibular impact

1. Introduction

High kinetic energy impacts to the skull are often associated with hemorrhagic brain and meningeal injuries whose mechanism has been examined in several studies and for various types of cranial impacts [1–5]. Boxers are subjected to craniofacial impacts, and even low-velocity impacts may be associated with major brain injuries because of the evident anatomical proximity between the face and the brain. Inertia effects may be observed when the head is violently shaken without a direct impact on the cranium (particularly in hyperextension) [3,6–8]. Movement of the head caused by a blow to the face can in itself cause direct concussion even if no anatomical injuries can immediately be observed, because the damage is axonal [9–12]. In literature, many authors demonstrated that the axon functional role can be altered even if it is not cut [11–14]. The brainstem pivots upon facial impact and suffers alterations by the subsequent shearing mechanisms [15]. Without fracture, the skull movement at impact may still have caused a direct brain contusion. Death may thus be secondary to force transmission to the brain, either by a so-called reflex mechanism that involves nerve conduction (by vagus nerve overstimulation) [16], or by central nervous system injury (axonal damage) [9,11,17].

When injuries cannot be identified by gross examination or histologically because of rapid death [18], the forensic and also sometimes clinical problem is how to relate facial trauma to injuries that led to an altered state of consciousness or even to sudden death. This issue may arise in criminal proceedings concerning voluntary acts of violence or involuntary events such as road accidents, and when no focal cerebral hemorrhage is identified although the victim died during the violence and no other cause of death has been identified. The digital approach is a means of attempting to understand mechanisms of injury arising from head/neck dynamics and also of simulating an event described by a third party. In this study, we firstly developed a finite element model of the head and neck in order to understand facial fracture mechanisms and to observe force transmission in the unit formed by the brain, brainstem and cervical spinal cord.

Secondly, we examined some examples of mandibular facial trauma in order to study the potential mechanisms of injury.

2. Materials and methods

2.1. Characteristics of the model

The finite element model that we developed is a combination of a head model developed by the Laboratory of Applied Biomechanics (LBA) and a neck model developed in collaboration with the École Polytechnique Montréal (EPM) (Fig.1).

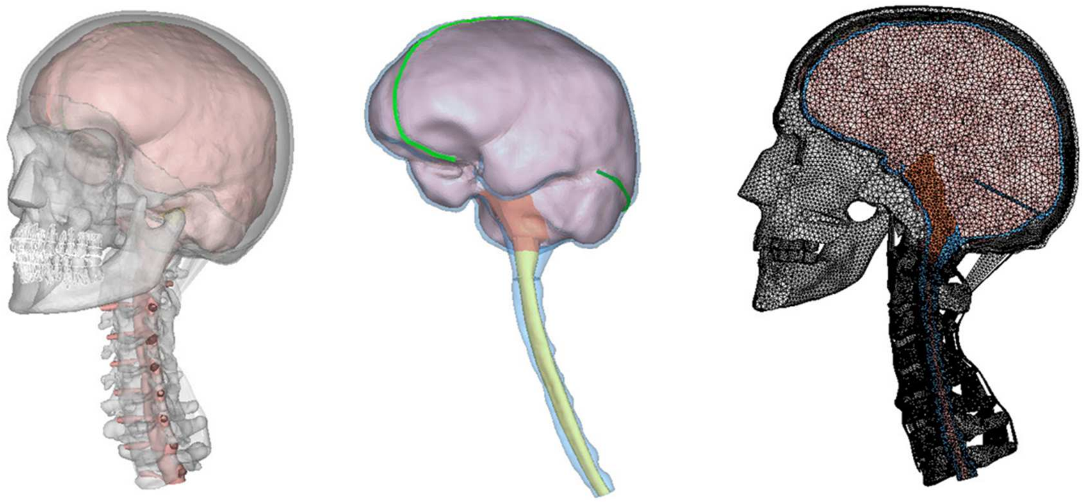

Fig 1: Finite element model of human head: a) whole model b) brain and spinal cord c) mid-saggital view.

Skull geometry was reconstructed from 1 mm Computerized Tomography (CT) scan slices of a 30-year-old man using MICMICS 12.3® software (Matérialise, Louvain, Belgium). The model was developed using Hypermesh® and Hypercrash® software (Altair Engineering, Inc., Detroit, MI, USA). The average size of the elements was 2 mm. The junction of the brainstem and the spinal cord was modeled in continuation of brainstem elements represented by tetrahedral elements. In the neck model developed by our laboratory (LBA, IFSTTAR in collaboration with ILABSPINE) [18,19], only the elements of the vertebral bodies and the spinal cord were retained.

Because of the anatomical complexity of the brain, the meninges and the neck, each part (pia mater, dura mater, falx cerebri, tentorium cerebelli, hemispheres, cerebellum, brainstem, cervical spinal cord, cervical vertebrae and ligaments) was modeled according to its different mechanical properties. Continuous meshing was used.

The skull was reconstructed in three layers representing compact bone (external and internal tables) and cancellous bone (diploë) modeled using tetrahedral elements. The brain and subarachnoid space, comprised between the brain and the skull to simulate the cerebrospinal fluid, were also modeled using tetrahedral elements. The dura mater, falx cerebri and tentorium cerebelli were modeled with three-node shell elements. The cerebellum and brainstem were individualized. A boundary condition was applied on the C7 vertebrae (rotation and translation were fixed).

2.2. Validation of the model

The model underwent several evaluations: four different sources of validation (the studies of Nahum et al., Trosseille et al., and Viano et al. (Fig. 2) and our own experimental studies in the laboratory) based on the most relevant experiments in the literature and according to three different configurations (three mandibular impacts: uppercut (Fig. 3), hook (Fig. 4) and anteroposterior impact (Fig. 5)) in order to study the influence of point of impact at the level of the mandible.

1. Nahum et al.[2] (test 37): blow with a rigid cylindrical bar (mass 5.59 kg, impact velocity 9.94 m/s) on the frontal region of a seated post-mortem human subject (PMHS), with the torso supported. The blow was delivered in a sagittal plane and an

anteroposterior direction, with the subject's head inclined forward at 45° in the Frankfort plane (Fig. 2).

2. Trosseille et al. [20]: blow with a rigid iron bar (mass 23.4 kg, impact velocity 7 m/s) to the face of a seated PMHS in an anteroposterior direction (experiment MS 428-2) (Fig. 2). In order to evaluate the response of our model, we compared the force of impact at the level of the frontal region and acceleration of the head at the center of gravity using Nahum's tests. Pressures at the frontal and occipital regions were measured and evaluated according to the tests of Nahum and of Trosseille.
3. Viano et al. [21]: impacts on 3 mandibular areas with different impact velocities with a hand mass of 1.67 kg: jaw (9 2 m/s), hook (11 m/s) and uppercut (6.7 m/s) (Fig. 2).
4. Experiments carried out in our laboratory in 2015: blow with a rigid cylinder (mass 5 kg, impact velocity 5 m/s) on a PMHS in order to evaluate the force of the blow at the mandibular symphysis and to assess the type of fracture observed [22].

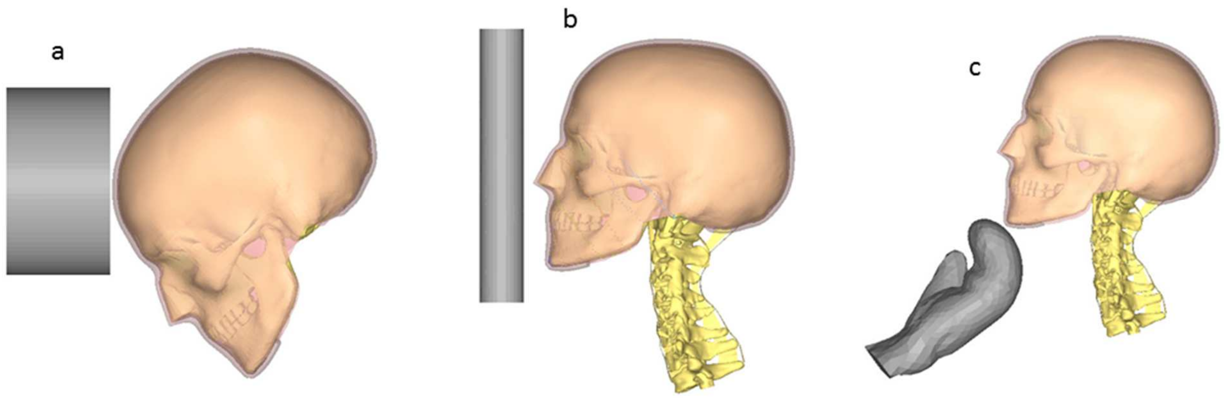

Fig 2 : Three numerical validation : Nahum et al (a), Trosseille et al (b), Viano et al (c).

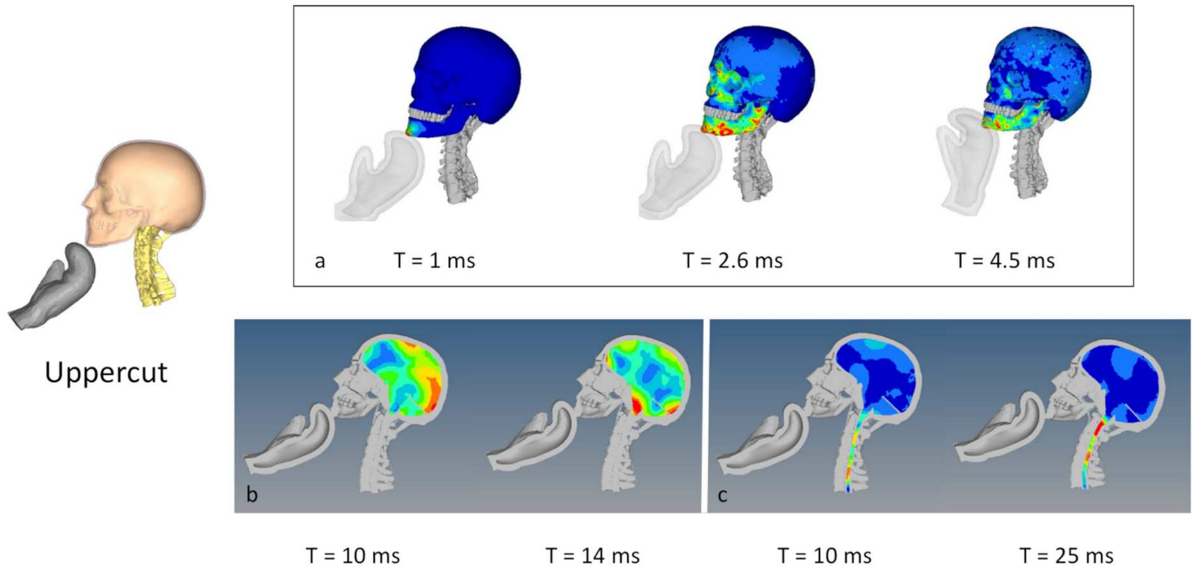

Fig 3: Stress propagation and distribution (Von Mises stress) in human head for uppercut scenario with focus on skull (a), brain (b) and cervical spinal cord (c).

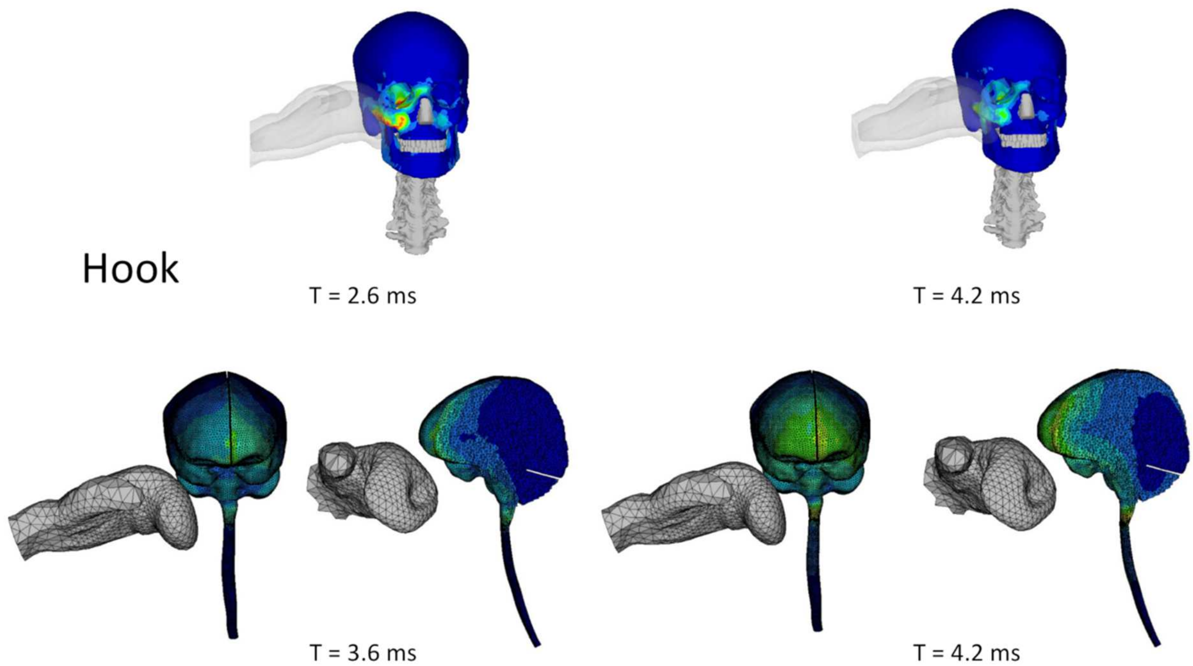

Fig 4: Stress propagation in human head for hook scenario with focus on skull and brain.

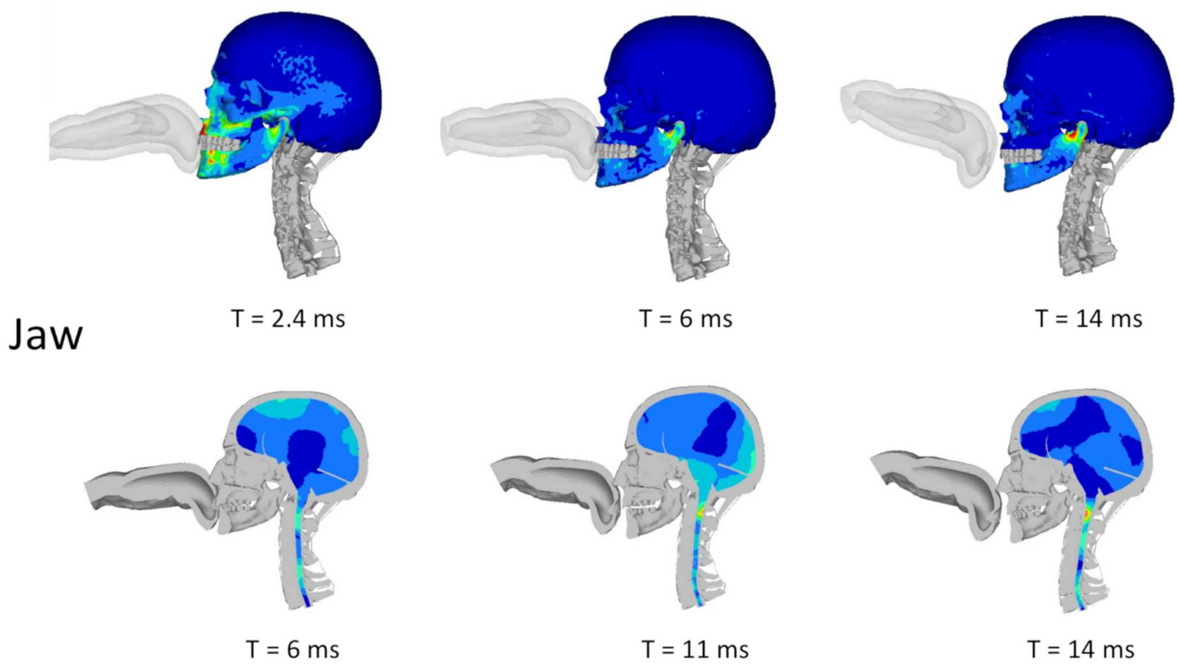

Fig 5: Stress propagation in human head for Jaw scenario with focus on skull and brain.

2.3. Resulting injuries

The injuries were examined at two levels: we observed firstly the distribution and propagation of stresses in the head, and secondly we evaluated the extension of the cervical spinal cord by measuring the space between the pons/medulla oblongata junction and C3.

The influence of impact location (uppercut, hook, face) on the development of stresses to the different parts of the brain (cerebrum, brainstem, cerebellum) and to the brainstem/spinal cord junction was also evaluated based on the tests of Viano et al. [18].

3. Results

The finite element model of the head consisted of 687,000 tetrahedral and hexahedral 3D elements and 85,000 shell elements (3 and 4 nodes). The mechanical parameters attributed to each anatomical part were based on the data of the literature (Table 1).

The finite element model of the neck consisted of 368,000 tetrahedral and hexahedral 3D elements and 122,000 shell elements (3 and 4 nodes). The head and neck together had a mass of 5580 grams.

We validated our digital model by reproducing the tests of Nahum, Trosseille and Viano. The comparative tests are summarized in the following table 2.

Anatomical components	Properties of the materials	Thickness (mm)	Density ρ (kg.m^{-3})	Young's modulus E (MPa)	Poisson coefficient ν	σ c: compressive yield stress	σ t: tensile yield stress	σ max	Ref.
Cancellous bone	elastoplastic		1500	4600	0.05	35 MPa	35 MPa		[23]
Cortical bone	elastoplastic	1.5	1900	15000	0.21	145 MPa	90 MPa		[23]
Mandible	elastoplastic	1.5	2500	13000	0.3				[24]
Teeth: enamel surface crown		0.5	1800	41000	0.3			220	[24]
Dentin density/alveolar area				18600	0.31				[24]
Scalp	elastic	5	1000	16.7	0.42				[23]
CSF	elastic		1040	0.12	0.49				[23]
Face	elastic	1 to 3	2500	5000	0.23			55	[23]
Brain	viscoelastic		1200	1225					[23]
Disc of the mandibular joint			1050	44.1	0.4				[25]
Dura mater	elastic	0.5		5	0.45				[26]
Pia mater	elastic	0.1		2.3	0.45				[26]
Vertebral cortical bone	elastoplastic	0.37-0.9		3319	0.3				[19]

Table 1. Mechanical properties of the anatomical elements of the finite element model

Authors	Impact site	Impactor mass (kg)	Impact velocity (m/s)	Forces (N)		Pressure (MPa)		Acceleration (m/s^2)	
				Literature	Digital simulations	Literature	Digital simulations	Literature	Digital simulations
Nahum et al. (1977)	Anteroposterior	5.59	9.94	8000N	9800N	Frontal 0.12 MPa Posterior fossa 0.08 MPa	Frontal 0.14 MPa Posterior fossa 0.06 MPa	2034	2019
Trosseille et al. (1992)	Anteroposterior	23.4	7	X	X	Frontal 0.09 MPa Occipital 0.018 MPa	Frontal 0.12MPa Occipital 0.011 MPa	X	X
Viano et al. (2005)	Anteroposterior Jaw Uppercut Hook	1.67	9.2 6.7 11	2349 SD 962N 1546 SD 857N 4405 SD 2318N	4300N 2300N 5080N	X	X	X	X
Experimental test (LBA)	Uppercut	5	5	3150 ± 1141 N	2600N	X	X	X	X

Table 2: Conditions of evaluation of the model in relation to the literature

The results obtained (Table 2) were in agreement with the data of the literature concerning impact forces, head acceleration (Nahum's tests) and the different pressures measured in brain tissue.

3.1. Variation of impact location based on Viano's tests

- Uppercut (Fig. 3)

Forces were propagated in an anterior to posterior direction, were greatest at the junction of the brainstem and the spinal cord, and were associated with strong stresses at the occiput with a backlash effect that was visualized by pressures at the frontal level. Stresses were distributed at the level of the mandible extending to the condyles and then to the base of the skull, and were also distributed to the cheekbone. Forces spread to the frontal bone, ethmoid bone and nasal bone. The force of the impact caused considerable cervical extension that led to strong stresses at the level of the cervical spinal cord and the brainstem. Major stresses were visualized at the origin of the mandibular symphysis without condylar fracture. Extension of the cervical spinal cord was 2.6 mm.

- Hook (Fig. 4)

Simulations carried out on the finite element model revealed no fracture. The extension observed was very small (0.1 mm). Forces were propagated from the punch at the temporomandibular joint to the opposite side at the temporal level, with stresses mainly exerted at the junction of the brainstem and the spinal cord, frontal regions and at the falx cerebri.

- Anteroposterior (Fig. 5)

Simulations carried out on the finite element model in the anteroposterior direction on the jaw revealed that forces were propagated from the incisors to the maxillary arches, the septum, the ethmoid bone and then towards the palate and the sphenoid bone. These stresses were distributed in the frontal and temporal regions up to the occiput and the foramen magnum. In brain tissue, stresses were propagated from the frontotemporal lobe to the occipital lobe, with greatest stress at the junction of the brainstem and the cervical spinal cord. A minimal fracture of the mandibular symphysis was visualized, with no penetrating or displaced fracture. Elongation of the cervical spinal cord was 2 mm, indicating cervical extension.

4. Discussion

4.1. Finite element model

Ours is one of the first complete models to include the fully modeled face and jaw, to which we added a neck model validated by the Laboratory of Applied Biomechanics in collaboration with the École Polytechnique de Montréal. Biomechanical studies have addressed mechanisms of injury either of the skull or of the spine, but not of the head and neck as a whole [18,19,23,27]. The relevance of this study is that it associates the cranial and cervical parts, which functionally are totally inseparable, in order to study their dynamics in facial impacts. Moreover, addition of the neck to our initial model allowed us to visualize the stresses exerted on the brainstem and the cervical spinal cord through three different mandibular impacts.

4.2. Force transmission and influence of impact location

Visible craniocerebral injuries (fractures, brain hemorrhages, contusions) give us little difficulty in understanding the mechanism involved in force transmission in a craniofacial impact. Our study also examines non-visible injuries, that is, non-hemorrhagic axonal injuries responsible for altered neurological functions and leading to death.

In anteroposterior facial impacts such as uppercuts, we observed hyperextension of the spinal cord/brainstem junction together with major stresses in this area, but without the cranial fractures, around the foramen magnum in particular, that have sometimes been observed in other studies [6,8,17,28]. Forces were distributed along the mandible to the base of the cranium, with stresses passing from the frontal lobes to the occipital lobes, associated with high pressure at the brainstem and along the spinal cord. In order to keep as close as possible to real-life situations, we carried out impacts of the types of punches received in boxing.

A hook-type punch, on the other hand, did not cause hyperextension of the spinal cord through their mechanism, but considerable force was propagated without decreased intensity of the stresses measured at the impact zone of the brainstem.

Previously published head and neck models only examined brain injuries, and did not address the dynamics and stresses of brain/spinal cord tissue of the head and neck as a whole. During impacts or falls on the chin, injury by elongation or even rupture of the cervical spinal cord have in fact been described in the literature but not measured [3,6–8,29,30]. Voigt et al. reported brainstem injuries produced not only by hyperextension or anteflexion but also by torsion or other forces applied to the head [31]. Depending on the type of accident, brainstem injuries (partial severance) have been reported in vehicle drivers or passengers in high-velocity impacts where the face or forehead hit the dashboard or windscreen.

Our findings after hook-type punches were in agreement with those of Zivković et al. [32]. The location of cranial impact associated with specific cranial fractures is predictive of the presence or absence of pontomedullary injury. Lateral and frontal impacts are associated with the absence of pontomedullary injury, whereas impacts on the chin and the absence of direct cranial trauma are associated with pontomedullary injury, as we confirmed in our study. Jaw impacts cause violent movement of the head responsible for immediate craniocervical dislocations that may cause indirect brainstem injury, generally pontomedullary, because the pontomedullary junction is the thinnest, and therefore the weakest, anatomic part of the brainstem [32]. As we described in our previous publication [22], during a mandibular impact kinetic energy is transmitted from the mandible to the temporomandibular joints and then to the base of the cranium and to the brain. In our tests, we did not observe any fracture of the skull base around the foramen magnum because of the lower impact velocities and forces used. In both situations, transmission of the impact force was decreased. The energy would thus be sufficient to produce a pontomedullary injury, but not sufficient to produce a fracture of the skull base. The development of pontomedullary injuries is dependent on impact energy and also on the position of the fracture, and less dependent on head movement. Mandibular and facial impacts cause acceleration of the head and rotational acceleration of the brain, and with sufficient impact energy, they may lead to rotation and deformation of the brain responsible for the brainstem injuries that we identified. The brainstem is not only a central sensory-motor pathway, but also a regulatory center for the major vegetative functions: vigilance, heart rate, respiratory rate, in particular at the level of the medulla oblongata. Simple contusion or compression of the medulla oblongata can thus lead to loss of consciousness and vegetative dysregulation that may result in death.

4.3. Protection of the face, fractures and anthropometrics

Although there have been controversial findings on facial fractures and brain injuries [33–36], our previous study and that of Zivković et al. [22,32] demonstrate the role of the facial bones in absorbing energy, protecting the brain and brainstem from the transmission of high kinetic energies. Mandibular fractures occur essentially at the impact point of each location, thus decreasing by half the force transmitted to the brain, as shown in our first article [22] in frontal and uppercut impacts. In lateral impacts, forces are transmitted directly to the base of the skull and so to the brain, without extension of the spinal cord but without decrease of stresses at the junction of the brainstem and the spinal cord. This mechanism of energy transmission has been described by Tse et al., by Zandi and Seyed Hoseini and by Lee et al., [5,37,38]. Tse et al. stated that the facial fractures closest to the brain are a major risk factor for underlying brain injuries [4].

In the literature, numerous comparative clinical studies have examined the number of fractures in relation to the severity of the brain injuries observed and have established a correlation, but without analyzing the dynamics of the forces applied nor their velocities.

These descriptive studies do not take into account the unique characteristics of each individual and the multiple factors that intervene in the mechanism of injury: shape of the mandible, bone density, dentition, underlying disease conditions, age… Finite element modeling offers an alternative to experimental research, enabling the digital reproduction of situations of injury and the possibility of evaluating an infinite number of conditions and injuries.

4.4. Limitations

Our modeling of a skull, brain and its spinal cord made it possible to locate stresses that had no clinical consequences. However, it could not reproduce axonal injuries and the extremely complex brain interconnections of the various sensory-motor pathways. Differentiation of gray and white matter and fluid-structure modeling of the CSF and brain vessels need to be added to improve this model.

4.5. Future prospects

Understanding the influence of the mode of impact and of the victim's characteristics on the development of brain lesions is of major importance. But the unique nature of the impact is not the only factor that has to be taken into account. So in the light of these data, we hope to study the effect on brain tissue of multiple lower-energy impacts and also to modify the environment by evaluating the stress produced when the facial impact occurs when the individual is lying on the ground or immobilized against a hard surface.

5. Conclusion

Using digital simulations based on a finite element model of the head and neck, we were able to analyze extra- and intracranial injuries following different mandibular impacts. This digital study enabled us to confirm the involvement and extension of the brainstem and cervical spinal cord during low-velocity impacts. Deaths secondary to this type of impact without identifiable histological lesions may be related to excessive stressing of the brainstem, along which pass sensory-motor pathways and the vagus nerve, and which is a regulatory center for the major vegetative functions. Biological parameters are different in each individual, and by using digital modeling they can be modulated at will (mandible shape, dentition…) for a realistic approach to forensic applications.

References

[1] T.J. Walilko, D.C. Viano, C.A. Bir, Biomechanics of the head for Olympic boxer punches to the face, Br. J. Sports Med. 39 (2005) 710–719. https://doi.org/10.1136/bjsm.2004.014126.

[2] A.M. Nahum, R. Smith, C.C. Ward, Intracranial Pressure Dynamics During Head Impact, SAE International, Warrendale, PA, 1977. https://doi.org/10.4271/770922.

[3] H. Gunji, I. Mizusawa, K. Hiraiwa, The mechanism underlying the occurrence of traumatic brainstem lesions in victims of traffic accidents, Leg. Med. Tokyo Jpn. 4 (2002) 84–89.

[4] K.M. Tse, L.B. Tan, S.J. Lee, S.P. Lim, H.P. Lee, Investigation of the relationship between facial injuries and traumatic brain injuries using a realistic subject-specific finite element head model, Accid. Anal. Prev. 79 (2015) 13–32. https://doi.org/10.1016/j.aap.2015.03.012.

[5] K.M. Tse, L.B. Tan, S.J. Lee, S.P. Lim, H.P. Lee, Development and validation of two subject-specific finite element models of human head against three cadaveric experiments, Int. J. Numer. Methods Biomed. Eng. 30 (2014) 397–415. https://doi.org/10.1002/cnm.2609.

[6] R. Lindenberg, E. Freytag, Brainstem lesions characteristic of traumatic hyperextension of the head, Arch. Pathol. 90 (1970) 509–515.

[7] T. Kondo, K. Saito, J. Nishigami, T. Ohshima, Fatal injuries of the brain stem and/or upper cervical spinal cord in traffic accidents: nine autopsy cases, Sci. Justice J. Forensic Sci. Soc. 35 (1995) 197–201. https://doi.org/10.1016/S1355-0306(95)72661-2.

[8] J.E. Leestma, M.B. Kalelkar, S. Teas, Ponto-medullary avulsion associated with cervical hyperextension, Acta Neurochir. Suppl. (Wien). 32 (1983) 69–73.

[9] T.A. Gennarelli, L.E. Thibault, J.H. Adams, D.I. Graham, C.J. Thompson, R.P. Marcincin, Diffuse axonal injury and traumatic coma in the primate, Ann. Neurol. 12 (1982) 564–574. https://doi.org/10.1002/ana.410120611.

[10] G. Davidoff, M. Jakubowski, D. Thomas, M. Alpert, The spectrum of closed-head injuries in facial trauma victims: incidence and impact, Ann. Emerg. Med. 17 (1988) 6–9.

[11] J. Sahuquillo, M.A. Poca, Diffuse axonal injury after head trauma. A review, Adv. Tech. Stand. Neurosurg. 27 (2002) 23–86.

[12] A. Büki, J.T. Povlishock, All roads lead to disconnection?--Traumatic axonal injury revisited, Acta Neurochir. (Wien). 148 (2006) 181–193; discussion 193-194. https://doi.org/10.1007/s00701-005-0674-4.

[13] N. Besenski, D. Jadro-Santel, N. Grcević, Patterns of lesions of corpus callosum in inner cerebral trauma visualized by computed tomography, Neuroradiology. 34 (1992) 126–130.

[14] J.T. Povlishock, C.W. Christman, The pathobiology of traumatically induced axonal injury in animals and humans: a review of current thoughts, J. Neurotrauma. 12 (1995) 555–564. https://doi.org/10.1089/neu.1995.12.555.

[15] G. Belingardi, G. Chiandussi, I. Gaviglio, Development and validation of a new finite element model of human head, in: 2005.

[16] B. Schrag, P. Vaucher, M.D. Bollmann, P. Mangin, Death caused by cardioinhibitory reflex cardiac arrest—A systematic review of cases, Forensic Sci. Int. 207 (2011) 77–83. https://doi.org/10.1016/j.forsciint.2010.09.010.

[17] W. Ezzat, L.C. Ang, J. Nyssen, Pontomedullary rent. A specific type of primary brainstem traumatic injury, Am. J. Forensic Med. Pathol. 16 (1995) 336–339.

[18] M. El-Rich, P.-J. Arnoux, E. Wagnac, C. Brunet, C.-E. Aubin, Finite element investigation of the loading rate effect on the spinal load-sharing changes under impact conditions, J. Biomech. 42 (2009) 1252–1262. https://doi.org/10.1016/j.jbiomech.2009.03.036.

[19] E. Wagnac, P.-J. Arnoux, A. Garo, C.-E. Aubin, Finite element analysis of the influence of loading rate on a model of the full lumbar spine under dynamic loading conditions, Med. Biol. Eng. Comput. 50 (2012) 903–915. https://doi.org/10.1007/s11517-012-0908-6.

[20] X. Trosseille, C. Tarriére, F. Lavaste, F. Guillon, A. Domont, Development of a F.E.M. of the Human Head According to a Specific Test Protocol, SAE International, Warrendale, PA, 1992. https://doi.org/10.4271/922527.

[21] D.C. Viano, I.R. Casson, E.J. Pellman, C.A. Bir, L. Zhang, D.C. Sherman, M.A. Boitano, Concussion in professional football: comparison with boxing head impacts--part 10, Neurosurgery. 57 (2005) 1154–1172; discussion 1154-1172.

[22] L. Tuchtan, M.-D. Piercecchi-Marti, C. Bartoli, D. Boisclair, P. Adalian, G. Léonetti, M. Behr, L. Thollon, Forces transmission to the skull in case of mandibular impact, Forensic Sci. Int. 252 (2015) 22–28. https://doi.org/10.1016/j.forsciint.2015.04.017.

[23] H.-S. Kang, R. Willinger, B.M. Diaw, B. Chinn, Validation of a 3D Anatomic Human Head Model and Replication of Head Impact in Motorcycle Accident by Finite Element Modeling, SAE International, Warrendale, PA, 1997. https://doi.org/10.4271/973339.

[24] T.P. Bezerra, F.I. Silva Junior, H.C. Scarparo, F.W.G. Costa, E.C. Studart-Soares, Do erupted third molars weaken the mandibular angle after trauma to the chin region? A 3D finite element study, Int. J. Oral Maxillofac. Surg. 42 (2013) 474–480. https://doi.org/10.1016/j.ijom.2012.10.009.

[25] E. Tanaka, T. van Eijden, Biomechanical behavior of the temporomandibular joint disc, Crit. Rev. Oral Biol. Med. Off. Publ. Am. Assoc. Oral Biol. 14 (2003) 138–150.

[26] C.Y. Greaves, M.S. Gadala, T.R. Oxland, A three-dimensional finite element model of the cervical spine with spinal cord: an investigation of three injury mechanisms, Ann. Biomed. Eng. 36 (2008) 396–405. https://doi.org/10.1007/s10439-008-9440-0.

[27] R. Willinger, H.S. Kang, B. Diaw, Three-dimensional human head finite-element model validation against two experimental impacts, Ann. Biomed. Eng. 27 (1999) 403–410.

[28] D.A. Simpson, P.C. Blumbergs, R.D. Cooter, M. Kilminster, A.J. McLean, G. Scott, Pontomedullary tears and other gross brainstem injuries after vehicular accidents, J. Trauma. 29 (1989) 1519–1525.

[29] S. Chatelin, C. Deck, F. Renard, S. Kremer, C. Heinrich, J.-P. Armspach, R. Willinger, Computation of axonal elongation in head trauma finite element simulation, J. Mech. Behav. Biomed. Mater. 4 (2011) 1905–1919. https://doi.org/10.1016/j.jmbbm.2011.06.007.

[30] D. Sahoo, C. Deck, R. Willinger, Brain injury tolerance limit based on computation of axonal strain, Accid. Anal. Prev. 92 (2016) 53–70. https://doi.org/10.1016/j.aap.2016.03.013.

[31] G.E. Voigt, G. Sköld, Ring fractures of the base of the skull, J. Trauma. 14 (1974) 494–505.

[32] V. Zivković, S. Nikolić, D. Babić, F. Juković, The significance of pontomedullar laceration in car occupants following frontal collisions: A retrospective autopsy study, Forensic Sci. Int. 202 (2010) 13–16. https://doi.org/10.1016/j.forsciint.2010.04.013.

[33] J.F. Kraus, T.M. Rice, C. Peek-Asa, D.L. McArthur, Facial trauma and the risk of intracranial injury in motorcycle riders, Ann. Emerg. Med. 41 (2003) 18–26. https://doi.org/10.1067/mem.2003.1.

[34] R. Gassner, T. Tuli, O. Hächl, A. Rudisch, H. Ulmer, Cranio-maxillofacial trauma: a 10 year review of 9,543 cases with 21,067 injuries, J. Cranio-Maxillo-Fac. Surg. Off. Publ. Eur. Assoc. Cranio-Maxillo-Fac. Surg. 31 (2003) 51–61.

[35] S.D. Nikolic, T.C. Atanasijevic, V.M. Popovic, M.V. Soc, The facial-bone fractures among fatally injured car occupants in frontal collisions, Leg. Med. Tokyo Jpn. 11 Suppl 1 (2009) S321-323. https://doi.org/10.1016/j.legalmed.2009.01.079.

[36] H.T. Keenan, S.I. Brundage, D.C. Thompson, R.V. Maier, F.P. Rivara, Does the face protect the brain? A case-control study of traumatic brain injury and facial fractures, Arch. Surg. Chic. Ill 1960. 134 (1999) 14–17.

[37] M. Zandi, S.R. Seyed Hoseini, The relationship between head injury and facial trauma: a case-control study, Oral Maxillofac. Surg. 17 (2013) 201–207. https://doi.org/10.1007/s10006-012-0368-z.

[38] K.C. Lee, S.-K. Chuang, S.B. Eisig, The Characteristics and Cost of Le Fort Fractures: A Review of 519 Cases From a Nationwide Sample, J. Oral Maxillofac. Surg. Off. J. Am. Assoc. Oral Maxillofac. Surg. (2019). https://doi.org/10.1016/j.joms.2019.01.060.