NEW CONCEPT OF CONTRECOUP LESION MECHANISM

Modal analysis of a finite element model of the head

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ABSTRACT

A theoretical modal analysis of a finite element model of the human head in its sagittal plane is proposed. The result applications to the crash biomechanics field are presented. For the model validation, this approach allowed us to determine the mechanical properties of the material used to model the subarachnoid space so as to obtain the first natural frequency and mode shape in accordance with the mechanical impedance recorded on the human head in vivo. The main result of our analysis is reached at the mode shape level and in particular with respect to its quasi insensitivity to mechanical properties of the materials involved. This 1st mode shape reveals a rotation of the brain mass around a point located at the center of a circle which describes the skull contour in the sagittal plane. This motion leads to significant brain stresses at the frontal lobes level and mainly at their interface with the orbital floor. Considering the response of a structure to a blow as a superposition of excited vibration modes, we show how occipital impacts lead to frontal injuries. This result is then illustrated in the temporal field by calculating the brain motions and intracerebral stresses under impact conditions.

INTRODUCTION

This study is part of a research into the assessment methods of head protection systems and finds applications in the development of vehicle or helmet approval tests. In general, the work, jointly carried out by INRETS and CNRS, was aimed at studying the injury mechanisms involved in case of head trauma.

The purpose of these studies is to determined the physical parameters which involve injuries and then to transpose the proposed mechanisms to test dummies through recording the key parameters under real impact conditions. To date, the parameter selected has been the acceleration of the gravity center of the head which cannot, in the general opinion - even if it allows to give a certain idea of the impact severity - express the risk of a specific injury which can be due to a relative motion between the brain and the skull, to a brain deformation, to a local bone deformation, to the displacement of a bone plate constituing the skull or to the propagation of a shock wave.

The work presented here addresses the mechanisms involved in case of contrecoup injuries. If the occurence of contrecoup injuries is not to be proved, the explanation of such a phenomenon and the process leading to it has not been clearly defined and to date has only been the subject of assumptions. Observation in epidemiological studies [1,2,3] reveals injuries in frontal impact conditions similar to those observed in occipital impact conditions. These injuries are located at the frontal lobes and front skull interface. In case of side impact, these studies show an injury symmetry at the front level of temporal lobes proving the occurence of contrecoup injuries in both impact situations.

In the literature, various attempts to explain such a phenomenon were made. Thus, in 1958, Gross [4] suggested that these injuries were produced by a cavitation phenomenon in the cerebral matter, but this result has not been demonstrated experimentally. Engin [5] explained the contrecoup injuries by the shock wave reflection, a result which was also purely theoretical and has been obtained on viscoelastic fluid models. In 1987, Gennarelli [6] showed that there were tensile stresses in the area located opposite to the impact point and ascribed these stresses to the appearance of contrecoup injuries.

All the proposals do not seem us very convincing, first because they all remain at a theoretical stage without any comparison against in vivo experiments, second they only partially explain the contrecoup injuries observed. Indeed, these proposals do not explain why contrecoup injuries are rarely observed in the occipital area in case of frontal impact. Then, it is difficult to admit that compression on the ipsilateral side and tension on the contralateral side involve injuries of similar types and localized in the same areas. Finally, the assumptions given below suggest intracerebral injuries while contrecoup injuries observed are generally focal and peripheral.

The new concept of contrecoup lesion mechanism presented in this paper is based on the vibration analysis of the human head in vivo, and then on modelling its dynamic behaviour using a head lumped model and then a finite element model. The results of the experiments carried out, the models obtained and the literature showed that the structure studied is made of visco-elastic solids of significantly different Young's moduli and Poisson's ratios. Despite damping phenomena, this structure is provided with an elastic vibration motion, which is more or less complex, after an impact.

The methods based on the head vibration analysis have often been used in the past [7,8], especially in the aim of drawing up head lumped models. The principle consists in submitting the structure to a not too violent blow and to record its response. The transfer fonction calculated from the data obtained allows the construction of a lumped model which in turn enables a theoretical study of the structure behaviour under real impact conditions.

Our study applies the same technique to a finite element model through a theoretical modal analysis of the model. The very principle of such an approach is basically different since a model with distributed parameters allows the distinction of medium-related deformations and possible relative displacements. The attraction of the modal analysis lies then at the model validation level which is an issue which has not yet or poorly been addressed for such a type of model, but also at the mode shape level which expressed the limit position of vibration motions for a prescribed natural frequency.

In the following section, we first detail the head vibration analysis and then present the theoretical models proposed in previous studies. We put the stress on the modal analysis of the finite element model which could explain the contrecoup injury mechanism. Finally, the mechanism concept is illustrated by an increment analysis in the temporal area of the head in case of frontal and occipital impact.

METHOD : HEAD VIBRATION ANALYSIS

The response of an elastic structure to an impact corresponds in the temporal field to a more or less complex vibration which is much longer than the impact duration, even if this structure has damping properties. This temporal response most often includes superimposed vibrations whose motion characteristics, amplitude, time and damping can be very different. The Fourier transform of the response expresses the structure behaviour in the frequency field and presents resonance peaks whose frequency, amplitude and damping are characteristic of a vibration mode.

To characterize a system, the response is related to excitation. In mechanics, the structure signature is often expressed by the Fourier transform of the velocity calculated from the recorded acceleration at a prescribed point of the structure, divided by the Fourier transform of the excitation force. This transfer function, called mechanical impedance, is a complex function of the frequency which enable the calculation of the structure response for any excitation type. In a given frequency range, the mechanical impedance clearly shows the natural frequencies corresponding to a specific vibration mode whose characteristics and effects on the whole structure behaviour are to be defined. These aspects of the vibration analysis have been detailed in the Harris work [9].

This first modelling step consisted in delivering blows to a human in vivo or in vitro head using a stricker equipped with a force transducer and in recording the structure response with an accelerometer placed near the impact area. In a second step, a mass-spring model has been developed to verify two conditions: the first one, which is evident, is that the theoretical impedance of the model is superimposed on the experimental impedance in the frequency range studied. The second verification, which is far more difficult to carry out, is the realistic interpretation of the model components and of its behaviour.

Even if they do not provide for the distinction between material deformation and relative displacements of the structure components, these lumped models were of a great help in understanding the head dynamic response when impacted. Indeed, this vibration analysis illustrated the mass behaviours in some frequency ranges, the natural frequencies and their damping, as well as mass "decoupling" phenomena followed by relative vibrations.

Various lumped models are proposed in the literature [7,8,10]. Most often, the head-related model is a system of two masses connected by a more or less complex set of springs and dampers. Interpreting these models and the injury mechanisms involved suggests that the frontal bone is placed in resonance with the other parts of the head and that the cerebral matter is thus submitted to a compression when the head is impacted.

For our part, we noticed that nearly all the impedance curves in the literature, as well as our own recordings [11], showed a natural frequency of about 120Hz in addition to the natural frequencies recorded at about 900Hz (see figure 1). These curves illustrate a mass behaviour up to the first natural frequency where a mass decoupling can be observed of the order of 1.5 kg. At the second natural frequency, a second mass decoupling is observed and beyond 900 Hz only a mass of about 0.8 kg still vibrates. These results led us to discard the two-mass principle as such a model could not explain the occurence of two natural frequencies in the structure response studied.

HEAD LUMPED MODEL

The head lumped model we proposed following the vibration analysis of in vivo and in vitro heads has been detailed in a recent study [12]. The main results obtained are given below.

The modal parameters relating to the first low frequency mode led us to distinguish the brain mass from the other masses involved and to develop a model which is represented in figure 2.

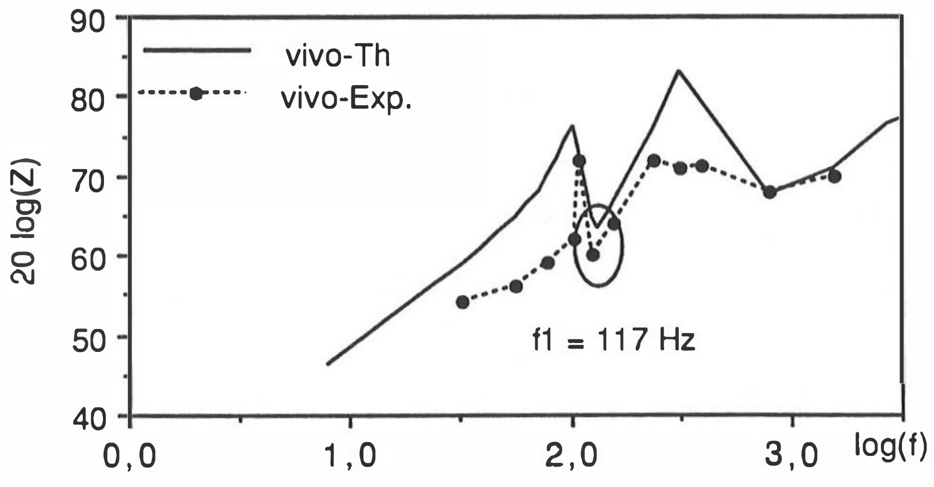

Figure 1: Experimental and theoretical curves of mechanical impedance of the head in vivo.

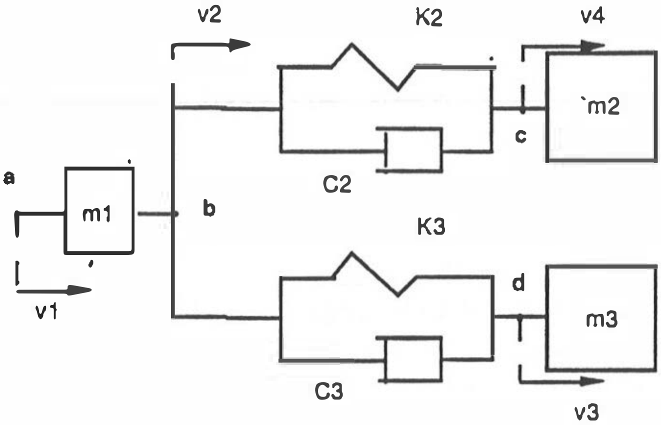

Figure 2: The lumped head model.

where :
m_t : total head mass,
m_1 : frontal bone mass,
m_2 : cerebral mass
$m_3 : m_t - (m_1 + m_2)$
$k_{2,3}$, $c_{2,3}$: stiffness, damping of m_2, m_3

This work showed that a relative motion exists between the brain and the skull beyond a 120 Hz frequency, and enables to calculate the relative motion characteristics as a function of the impact force and model parameters.

The lumped model let us to introduce a change in the injury mechanism as a function of impact energy distribution in the frequency range, and then to study the notion of tolerance limits in the frequency field [13].

Thus, for "long duration" impacts, when the impact spectrum is concentrated in a frequency range below 120Hz, the brain follows the head in its translation, vibration or rotation motion, leading to some intracerebral strains. When the impact amplitude or the amplitude of the head motions are significant, brain strains are significant too and all the conditions for the appearance of intracerebral injuries are thus observed. This is a typical phenomenon in an impacted subject wearing a helmet.

For "short duration" impacts, when the impact spectrum energy is distributed in a frequency range beyond 120Hz, the brain does not anymore follow the skull motions and a relative displacement can be observed. Of course, when the impact amplitude is significant, the relative displacement is significant too and peripheral cerebral injuries appear, such as sub-dural haematoma, compression of frontal lobes in their anterior part or alteration of frontal lobes at their interface with the orbital floor. This is a typical phenomenon observed in case of direct impact on a head not wearing a helmet.

In order to go beyond the limits of such a type of model as related to the brain intrinsic deformations, and to get more information on the vibration motion of the head at the first resonance frequency, we propose to extend the vibration analysis to the modal analysis of a distributed parameter model.

DISTRIBUTED PARAMETER MODEL

A lot of finite element models have been proposed in the past years and the main issue at stake concerning these models is their validation [14, 15, 16].

In a literature review on the finite element modelling of the head, Khalil and Viano [17] examined the non slip conditions between the skull and the brain. They studied the advantage of 3D models while 2D models which have not been rigorously validated were not fully applied, and they regretted that no model had never been compared against a living model under dynamic conditions. They concluded that, owing to the current development of the models, it was not possible to predict injuries or to define tolerance limits. Nevertheless, they considered that these models could be used to predict impact severity and to assess the efficiency of protection systems. We consider that even this last role cannot be fully ensured until the injury mechanisms are not known with certainty.

For our part, we present a 2D model calculated by finite element process and validated under dynamic conditions and in vivo analysis. To our knowledge, such an analysis has not been proposed in the literature. In addition to the model validation, we show how a theoretical modal analysis yields information on the injury mechanisms in relation with a prescribed vibration mode.

A simplified modelling of the brain-skull system and the interface conditions between them is contemplated. The study uses a two-dimensional and plane strain model in the sagittal plane. The brain mass is assimilated to a single deformable continuous medium, thus neglecting the influence of the membranes. This influence is much less important in the studdyed sagittal plane, than in the coronal plane [18]. The system has been discretized by 549 nodes and 573 elements as shown in figure 3. A perfect continuity was assumed between the thin layer modelling the subarachnoid space and the skull on the one hand and the brain on the other hand. The mechanical properties admitted for the brain and bone matters are respectively for an isotropic-elastic case [15,17,18] :

$$E_c = 675\ 10^3\ \text{Pa},\ \nu_c = 0.48 \qquad \text{and} \qquad E_o = 50\ 10^8\ \text{Pa},\ \nu_o = 0.2$$

The subarachnoid space is modelled by a material layer whose Poisson's ratio is of the order of ν_s =0.49 [17] and whose Young's modulus E_s is selected as a variable of the problem.

The assumed elasticity of this intersticial material seemed acceptable for the analysis proposed as all its damping characteristics have low effect on the value of natural frequencies and on mode shapes, given that this damping values are small in vivo (see fig. 1). This simplifying assumption does not impede neither the model validation, nor the opportunities of analyzing the injury mechanisms which stem from modal shapes. In a further stage, it will be nevertheless required to introduce subarachnoid space viscosity-related damping.

MODAL ANALYSIS OF THE DISTRIBUTED PARAMETER MODEL

After discretization of finite elements a modal analysis has been carried out to identify dynamic behaviour at low frequencies. At this stage of the finite element analysis of the head, we focussed on the main attraction of the modal analysis as validation technique but also to know the mode shapes that can be directly related to an injury mechanism as a function of the critical areas involved.

After discretization, the problem of free vibrations is written [19] :

$$[M]\ \{\ddot{u}(t)\} + [K]\ \{u(t)\} = \{0\} \qquad (1)$$

where $\{u\} = \{u_i,\ \ i=1$ à $n\}$ are generalized coordinates, [M] and [K] are the system mass and stiffness matrices.

Assuming an harmonic motion and putting $\{u(t)\} = \{[u]\ e^{i\omega t}\}$, the problem relates to the following eigenvalues equation :

$$[K]\ \{u\} = \lambda\ [M]\ \{u\} \qquad (2)$$

where $\lambda = \omega^2$: is the eigenvalue and ω its angular frequency, real if [K] and [M] are symmetrical.

With each eigenvalue λ_i, are associated a frequency f_i and an eigenvector $\{p_i\}$, characterizing a vibration mode.

Our study focussed on the first natural frequency f_1 of the system as this mode is conditioned by the presence of the brain mass and the stiffness of its interface with the skull. As the skull does not undergo strain at these frequencies, we selected for our analysis the outer boundary conditions considering that the nodes located at the skull border were encased. For the model validation, the value to be ascribed to E_s should be determined to obtain a model response which proposes a first mode at the 120Hz frequency, according to the experimental recording in vivo. In addition, as these experiments showed a "block" motion of a mass of the order of 1.5 kg, the mode shape must show a whole displacement of the brain mass whose rate of proper deformation remains low.

Figures 4 and 5 illustrate respectively the mode shapes and the change in the natural frequency value as a function of the subarachnoid material stiffness.

First, the results concerning the model including a subarachnoid layer of a stiffness similar to that of the brain matter has been analyzed, being thus close to the relative non slip conditions at the boundary between the brain and the skull. In this configuration, the first natural frequency is located at 3290 Hz. The value of this natural frequency is obviously a function of the elastic modulus of the brain matter, but this modulus should be brought to 10^3 Pa as compared to the 675 10^3 Pa proposed in the literature to obtain a first natural frequency at 115 Hz.

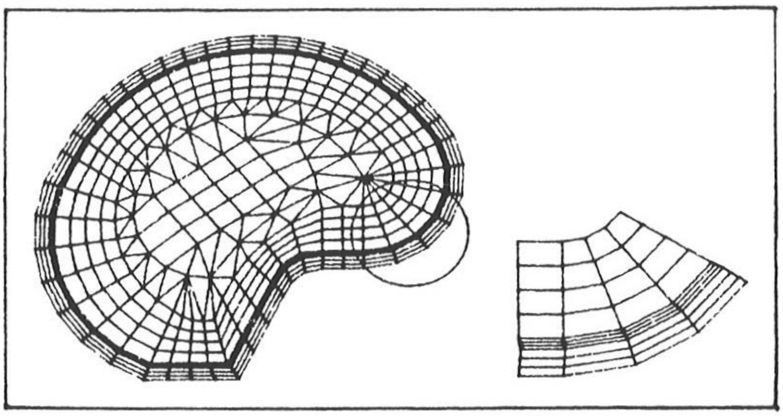

Figure 3 : Finite element model of the head in the sagittal plane. Observe detail for the subarachnoid space modelling.

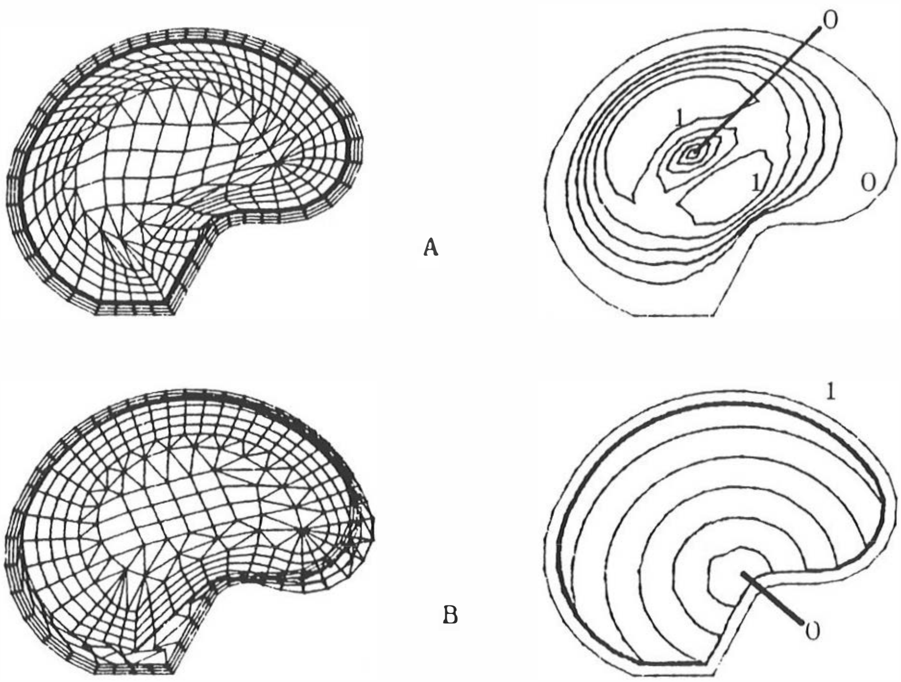

Figure 4 : Mode shape and isodisplacement curves relative to the first natural frequency of the head :

A) Es = Ec = $6.75\ 10^5$ Pa and f1 = 3290 Hz.

B) Es = 50 Pa and f1 = 150 Hz.

The mode shape is illustrated in figure 4A which shows the mode shape itself and its isodeformation curves. The displacement amplitude for this vibration mode is equal to zero in the brain central area, maximum around this area and decreases to zero at the skull interface. This figure shows the vibration of a rotating "brain ring" around a central node. The mass of this ring being less than that of the whole brain, this vibration mode is in contradiction with the modal parameters determined in vivo, as well as the 3290 Hz frequency.

When E_s is decreasing, the modal parameters of the first mode are changing too, and a decrease in the natural frequency value is recorded, associated with a progressive change in the mode shape from a ring vibration to a rotation vibration of the whole brain. This new mode shape is illustrated in Figure 4B by its shape and isodeformation curves for a value of E_s equal to 50 Pa, which sets the natural frequency to 150 Hz. In this configuration, the displacement amplitude is maximum at the uper border, it then decreases up to zero at the center of the circle described by the skull contour in the sagittal plane. This shape shows a rotation vibration of the whole brain, a motion that leads to few proper deformations of the brain.

Figure 5 shows that a natural frequency of 150 Hz is obtained for a value of E_s equal to 50 Pa. We also observed that this vibration mode in global rotation is steady for significant variations in the elastic modulus of the material used for modelling the subarachnoid space (E_s ranges from 25 Pa to 5 10^3 Pa). Furthemore, the quasi insensivity of the value of the natural frequency to significant variations of E_c reveals that this vibration mode actually produces few brain deformations. All these results confirm the assumption of a "block" motion of the brain mass as determined in experimental observations in vivo.

The consistency of these results, as well as their stability as a function of the mechanical parameters of the materials involved and which are more or less well approximated, allowed us to validate the model and to draw three basic consequences on the injury mechanisms in case of head trauma :

i) The relative displacements between the brain and the skull appear preferentially at the dome of the skull, which can lead to sub-dural haematoma located in this area in case of sudden rotation of the head as well as in case of sudden translation of the head in the sagittal plane. This result is nevertheless conditioned by the impact stiffness condition, as the 150 Hz frequency must be excited by the impact.

ii) In addition to the relative displacements, the mode shape presented in figure 4B leads to significant brain compressions at the frontal lobe level and their interface with the orbital floor, producing focal injuries often observed in this area.

iii) Due to the very nature of the mode shape relative to a mechanical vibration mode, the shape shown in figure 4B is only a vibration position resulting either from a frontal impact or an occipital impact.

The two last consequences stated above constitute a new concept of contrecoup injury mechanisms. In the following section, we will illustrate this concept by a temporal analysis of the model for frontal and then occipital impacts.

TEMPORAL ANALYSIS UNDER IMPACT CONDITION

This stage is aimed at showing the parametric temporal evolution for relative displacements and brain strains following a prescribed impact. The putting in equation of the finite element problem posed is performed according to the non damped motion equation which is written [19] :

$$[M]\,\{\ddot{u}(t)\} + [K]\,\{u(t)\} = \{F(t)\} \tag{3}$$

where {F(t)} is the excitation force vector.

The outer boundary conditions for this study are a free sliding condition along an anteroposterior axis (axis X) at the skull dome, orbital floor and skull base levels, complemented by an elastic support condition of a 500 Nm^{-1} stiffness at the occipital level. Our mathematical simulations have prove that this low connection stiffness does not affect the model behaviour when varying it from 100 to 1200Nm^{-1}.

The temporal load pulse applied along axis X is of triangular shape with a maximum amplitude of 100 N and a total duration of 10 10^{-3} s. If we assume that there is no deformation of the skull in the frequency range studied, duality between the frontal impact and the occipital impact is obtained by reversing the pulse sign which remains localized in the frontal area.

The parameters we selected to describe the structure behaviour with time are:

- displacement discrepancy along axis X, between the skull and the brain at the dome level, which constitutes a translation parameter noted "T".
- a parameter illustrating the brain rotation, given by the difference of displacement along axis X of a "brain point" located near the dome of the skull and a point of the brain stem base. The low influence of the brain proper deformations on the rotation parameter assessment has been proved by considering a frontal node and an occipital node in vertical motion. Rotation parameter "R" is expressed in meters to be superposed to the translation parameter. (the vertical distance between the points considered is 0.117m).
- tensile and compressive stresses along axis X in the frontal area (Sf) and the occipital area (So).

The relative motion of the brain with respect to the skull is first illustrated as a whole in figure 6 where the positions of the brain and the skull contour are represented at various instants following a frontal impact (fig. 6-F) and an occipital impact (fig. 6-O). This figure can be read in parallel with figure 7A which gives the evolution with time of parameters T and R for a frontal impact as well as the variations with time of R for an occipital impact. Variations of parameter T for an occipital impact are not illustrated in the figure, in the aim of simplifying it, but they can be easily deduced from the evolutions of T under frontal impact conditions.

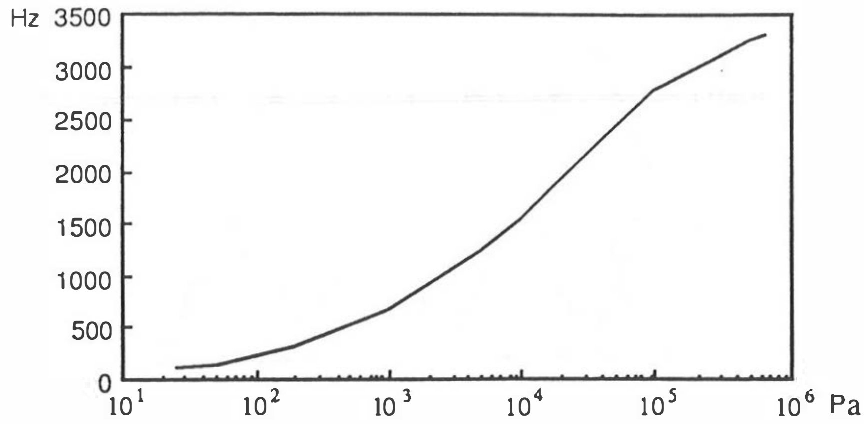

Figure 5 : Evolution of the first natural frequency of the head as a function of Young's modulus of the "subarachnoidal space material" Es.

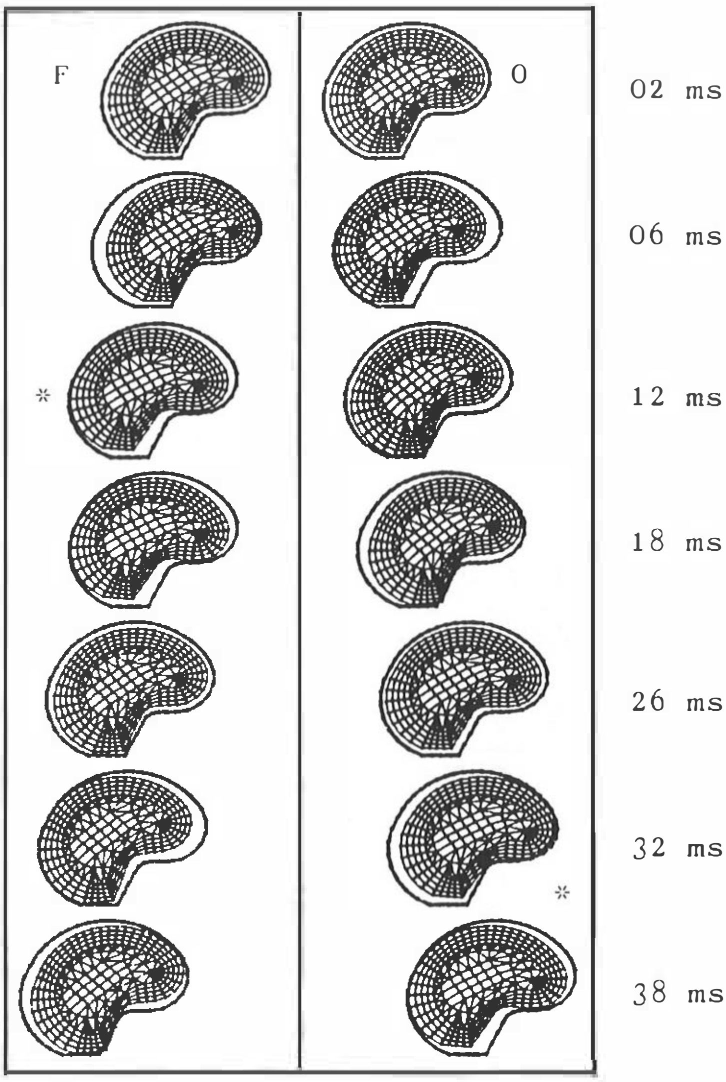

Figure 6 : Relative brain-skull motion in case of frontal F) and occipital O) impact. Skull and subarachnoïdal material is not drawn. Sckale:10.

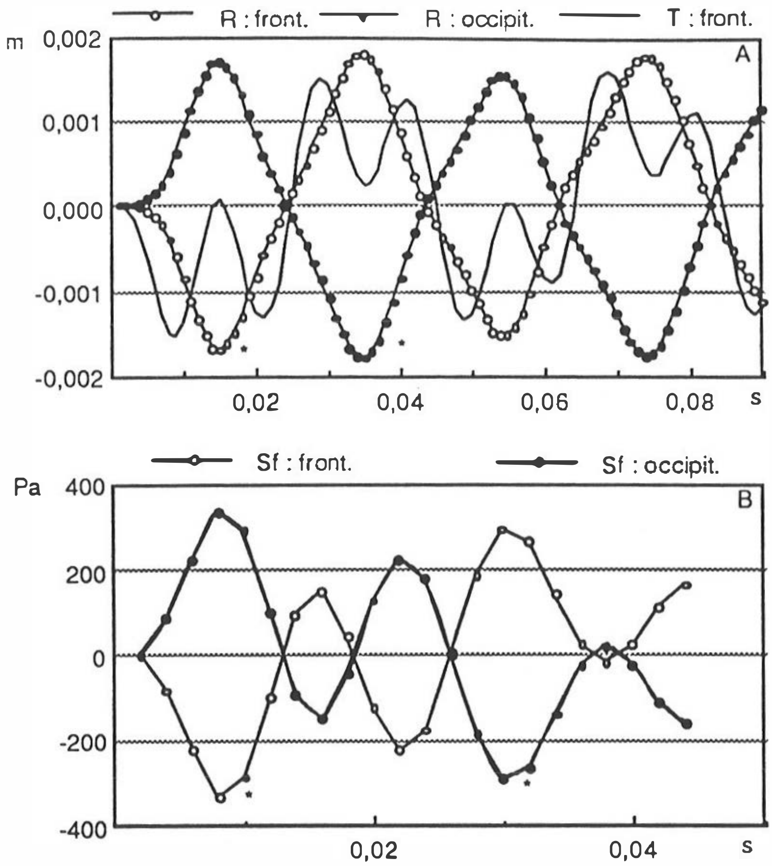

Figure 7 : Parameter evolutions as a function of time in case of frontal and occipital impact : A) Brain - skull relative motion
B) Intracerebral stresses.

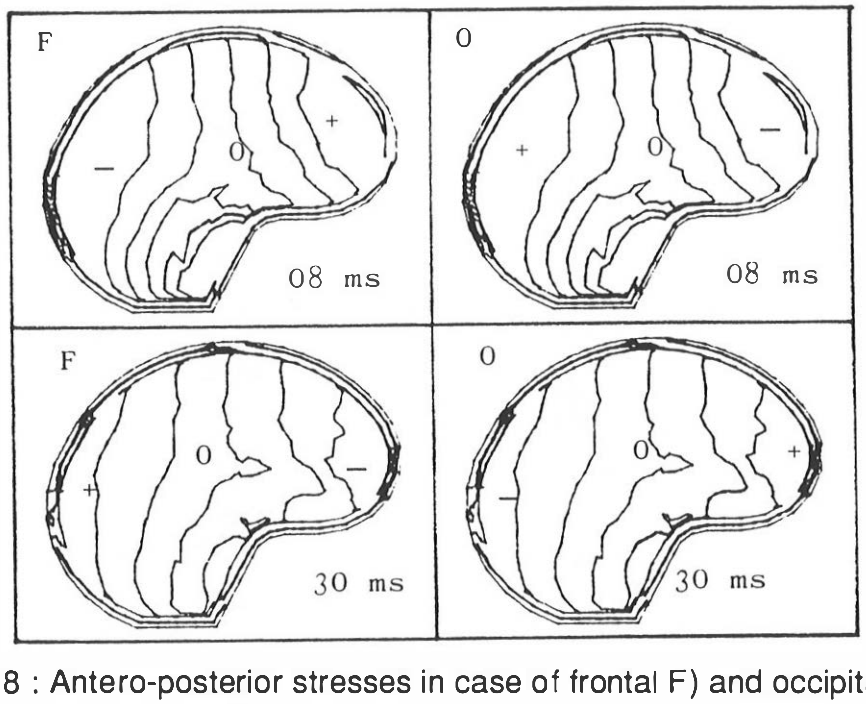

Figure 8 : Antero-posterior stresses in case of frontal F) and occipital O) impact, at critical time values. Isostress curves increment is 100 Pa, tension is positive and compression is negative.

In case of frontal impact, these figures show a forward translation of the brain with a maximum at 6 or 8ms and which is associated with a forward rotation motion reaching its extremum value at about 17 ms after the beginning of the impact. This kinematics produces significant compression stresses in the frontal area and tensile stresses in the occipital area (figures 7B and 8F), inducing a backward motion of the brain at about 30 ms, associated with a general backward rotation motion of the brain. This rotation reaches its extremum value at about 36 ms after the beginning of the impact. This configuration involves occipital compression stresses and frontal tensile stresses (figures 7B and 8F), resulting in a return to the previous condition. This vibration motion, which is theoretically perpetual, is in fact stopped after several beats due to damping phenomena.

For an occipital impact, the opposite motion is observed, i.e a backward displacement and rotation, followed by a forward motion similar to that described before, which can explain contrecoup injuries in the frontal area (figure 6-O, 7, 8-O).

The analogy between these two types of impact is illustrated by the * symbol written in figures 6 and 7 indicating that the critical stresses produced by a frontal impact are obtained with a delay of about 24 ms in case of an occipital impact. The absolute values of these parameters for both impact situations are not actually equal due to the vibration motion damping. These damping properties can explain the difference that seems to exist between head tolerances to frontal impact and to occipital impact.

The temporal analysis presented, based on modal analysis and lumped parameter modelling, allowed us to understand the occurence of identical injuries for frontal and occipital impacts. The contrecoup injury mechanism can be then attributed to a global vibration motion of the brain within the skull.

CONCLUSION

In this study, we applied the vibration analysis techniques used in conventional mechanics to the human head in order to characterize its dynamic behaviour under impact conditions. Such an analysis enabled the development of lumped parameter models of a spring-mass type and distributed parameter models of finite element type, validated by non destructive tests conducted in vivo on voluntary individuals. The head behaviour under real-world impact conditions was thus approximated by measuring the behaviour of these models under same impact conditions.

Constraints imposed by a lumped parameter model incited us to propose a finite element model whose mechanical properties have been adjusted by a modal analysis of this model. A natural frequency at 150 Hz and a mode shape characterized by a global brain rotation motion around a point located at the center of a circle described by the skull contour in its sagittal plane and corresponding to the first low frequency mode of the head, were obtained for a Young's modulus value of the material modelling the subarachnoid space of 50 Pa. The model validation was confirmed by a good stability of its modal behaviour, as it represents a mode shape which remained quasi unchanged for very significant variations of the Young's modulus of the subarachnoid space.

The modal behaviour demonstrated in the experiment, and for the two model types, illustrates the brain vibration motion after a frontal or occipital impact, as soon as the first mode frequency is excited. The very nature of this modal behaviour allowed us to describe the contrecoup injury mechanism through the vibration motion of the brain within the skull.

The time incremental analysis of the finite element model in the temporal field, for a frontal impact and then an occipital impact corroborated this contrecoup mechanism concept. Indeed, brain displacements and critical stresses determined for a frontal impact are similar for an occipital impact with a delay of the order of 24 ms. Damping, which has not been considered in our study, occurs at the vibration amplitude change but not at the vibration phenomenon itself. Damping properties of the materials consituting the head can thus explain the higher head tolerance to an occipital impact as compared to a frontal impact.

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