DETERMINATION OF CEREBRAL MOTION AT IMPACT
THROUGH MECHANICAL IMPEDANCE MEASUREMENT

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ABSTRACT

Cerebral injury mechanisms in translation configuration with and without impact find different origins in the literature. In order to better understand these mechanisms, a cerebral motion analysis through the "internal impedance" study of the brain is proposed.
For the test, a bovin brain is placed in a closed water-filled plastic box fixed on a metallic support. This support is connected to a spring and the sliding frictionless motion is carried out in the spring axis direction.
The impulse force is given to the support or to the box to simulate configurations with or without impact respectively. The input force, as well as the support, box and brain acceleration with time are recorded. A signal processing system gives the mechanical impedance.
For a rigid mass-spring system the theoretical impedance is
$z = iwm + k/iw$. In our test when the brain, box and support show displacements similar to those of a rigid mass then the three impedances recorded can be superimposed to the theoretical model.
The first test series included a translation impulse test without impact. The three impedance curves recorded are close to the theoretical curve up to 200 Hz. Above 200 Hz the two external impedances, i.e box and support, drop and can be superimposed to a new mass-spring system where the mass value is the precedent total mass less the brain mass. Above this critical frequency the internal impedance, i.e the brain, raises but this has no physical significance as there is only a little energy in the acceleration signal above 200 Hz.
A second test series dealt with the translation impulse with impact. In spite of a force impulse twenty times lower than the previous one due to the elasticity of the box, the internal impedance recorded is very close to those measured in the configuration without impact. This means that the brain moves like the box and the support and that it is stationary at higher frequency.
Mainly three conclusions can be drawn up at this stage of the study:
- the method seams to be accurate enough to predict the cerebral skull relative displacement.
- the brain does not seem to follow the box movement above 200 Hz and that despite a specific weight very close to the water one.
- to the exclusion of contact phenomena, no significant differences at

the cerebral motion level should be observed in configurations with or without impact.

INTRODUCTION

Our investigations aimed at better knowing the mechanisms of cerebral injuries in case of a rectilinear blow with or without impact and in case of a sudden rotation of the head. They allowed drawing up the human tolerance limits to blows. The development of anthropomorphic models used to assess the safety characterics of a vehicle in case of accident is an application of such studies.
Epidemiology, experiments on animals, and to a less extend experiments on cadavers gave us information data relating to the injury types observed in different blow configurations. Even when the kinematic values were known, these studies did not always allow defining the internal mechanisms that led to the injury, nor the values of the involved physical parameters. Different types of approach are given in the bibliography.
Physical models are often used to display the gel motions in a box subject to an impact. Thus it was shown that a sudden rotation of the head led to shearing stresses in the cerebral mass and at the skull/brain interface level.
The mathematical models of "viscoelastic solid" type are descriptive but they do not allow obtaining numerical values of the brain stress as the skull/brain limit conditions are not well known.
The mathematical models of fluid or viscous fluid type in a cylinder or a sphere well show the intracerebral shearing motions in case of head rotation as well as the shock wave propagation in case of translation with impact. The maximal values of angular micro-deformations proposed range from 0.02 to 0.05 and the brain-related theoretical models lead to head angular acceleration values of about $3.5 \cdot 10^3$ rd/s^2.
The mathematical models of mass-spring type, based on the frequency analysis of blows to cadavers or living subject appropriately consider the complexity of the skull-brain system but can lead to various interpretations. With this technique, the difficulty lies in knowing first what masses in relative motion are to be considered and then the physical nature of the cerebral mass in retention in the cerebrospinal fluid.
If all the cerebral injuries due to the sudden head rotation are unanimously attributed to the differential intracerebral and cranio-cerebral (1), (2) acceleration , according to the authors, those connected with the translation with impact are due to the maximal negative values of the pressure wave amplitude (3), to the simple compression of the cerebral mass involved by the displacement of the cranial bones (4), or to the sudden displacement of the cerebral mass in the skullcase (5).
In this paper, some basic answers to such questions are given with a study of the possible inertial motions of the cerebral mass in retention in a fluid.

MECHANICAL IMPEDANCE OF THE HEAD

As a first step, no physical or mathematical model has been used in our approach which aimed at studying the problem in all its complexity. The principle used was based on the black box identification through the study of the impulse response. In the studied case, the test carried out on a human cadaver or live human consisted in measuring the punctual impedance of the head. The difficulties met relate to the interpretation of the mass-spring models obtained from these transfer functions.

The frontal impedance curves of the head given in the bibliography show an antiresonance and a resonance ranging between 600 Hz and 1 000 Hz, with a behavior in mass at low and high frequencies. The associated mechanical model is a set of two masses (0.4 kg and around 4 kg), (6), (7), (8), linked between themselves by a damping spring connection (6.10^3 N/m, 160 Ns/m).

In his model interpretation, STALNAKER considers that the frontal bone vibrates and is placed in resonance with the other parts of the head leading to the cerebral mass compression. On the contrary, VIANO (9) thinks that it is the cerebral mass which vibrates with respect to the other parts of the head.

For our part we consider that the theoretical curve of this double mass model closely corresponds to the experiment impedance curve, but it does not take into account the anti-resonance and the resonance systematically observed between 100 and 200 Hz and which question the principle of the double mass model.

To further our understanding of this point, the impedance measurement was run on three human cadavers and on one live human subject (fig. 1). During the experiment, blows were delivered to the frontal bone using a stricker and an accelerometer. The test demonstrated an anti-resonance and a resonance at 100-200 Hz with a varying damping factor depending on whether the subject was live or dead.

These results led us to introduce a new model including a mass m1 in series with two "mass-spring-damper" systems placed in parallel (m2, k2, c2 and m3, k3, c3); see figure 2.

At first, the damping factor was disregarded , thus the head impedance equation yields:

$$Z = j\left(wm1 + \frac{wk2m2}{k2-w^2m2} + \frac{wk3m3}{k3-w^2m3}\right)$$

with the following approximate mass distribution:

m1 = 10%, m2 = 35%, m3 = about 55%

and a stiffness k3 about three times higher than the k2 stiffness.

The approximate determination of the damping factors sets the order of magnitude of values c2 and c3 to 100 and 800 Ns/m respectively. These two parameters will be more accurately determined further by plotting the impedance function according to Nyquist.

One of the possible interpretation is that the frontal bone of mass m1 vibrates with respect to two separate masses: cerebral mass m2 and mass of the other parts of the head m3. If this interpretation can be

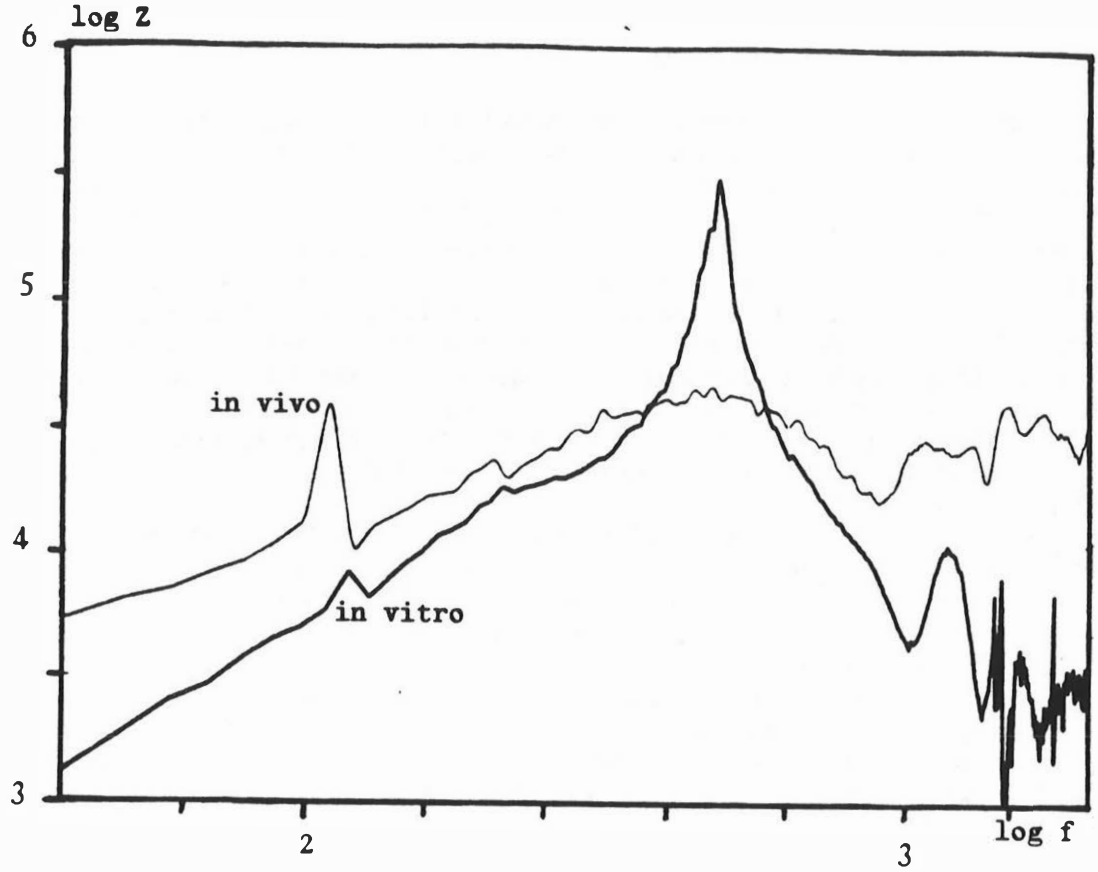


Fig. 1 : Frontal mechanical impedance recorded in vivo and in vitro

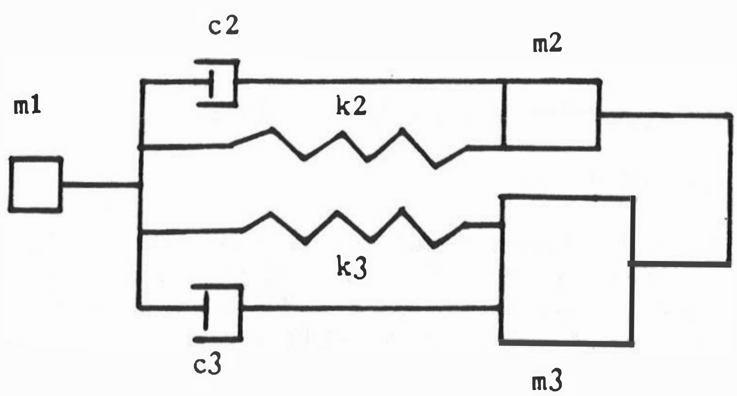


Fig. 2 : New mass-spring model of the head

validated, this model will allow calculating the skull/brain response to a blow and their relative motion. The recording of the head impedance could then constitute a test for system failure detection if used before and after a traumatic impact. Interesting research possibilities on the cerebral injury criteria could be offered through the analysis of parameters k2 and c2.
In order to check whether this interpretation of the model is plausible, the different possibilities of gravity motion of the cerebral mass will be studied in the following section.

METHOD

The method used to determine the inertial motion of the cerebral mass of density 1.04 in retention in a fluid of density 1.00 is based on the comparison of a closed water-filled box containing a brain with a rigid system of identical mass.
The tests consisted in placing a bovin brain into a closed flexible plastic box filled with water and firmly attached on a highly rigid support fixed to a spring. Displacements were carried out without friction in the horizontal plane and in the spring axis (fig.3).
An impulse was given to the box or to the support to respectively simulate a translation with or without impact. The striker force and the support, box or brain acceleration were simultenously recorded at a sampling frequency equal to 10 k Hz.
For a rigid body of mass M connected to a spring of stiffness K, displacement x of the mass was given by the Duhamel integral according to the natural frequency of system w_n and impact force F(t) given in the following equation:

$$x = \left(\cos wnt \int_0^t F(t) \sin wntdt + \sin wnt \int_0^t F(t) \cos wntdt\right) / (Mwn.)$$

As this calculation does not take into account the damping factors our theory-experiment comparison will be reduced to the first vibration.
The mechanical impedance of the system is expressed by $z = iwM + K/iw$. In our experiment, when the brain-box-support motion is similar to that of a rigid body, the three displacements calculated and the three impedances recorded can be superimposed to the theoretical model. Thus, any deviation with respect to displacement or theoretical impedance showed a deformation of the system or a relative motion of its components.
In this study three test series were run: a first one on the support only in order to check the validity of the method used. The two others illustrated the sudden translation configuration with and without impact as the impulse was first given to the support, and then to the box itself.

RESULTS AND DISCUSSION

The acceleration curves observed with time present diracs of 1 ms width and 60 to 80 g of amplitude when the accelerometer is fixed on the support and vibrations of +/- 5 g when the accelerometer is placed in the brain. At first they show that the behavior of the support-box-water-brain system is not similar to that of a rigid model. It is interesting to note that the brain acceleration curve does not significantly change

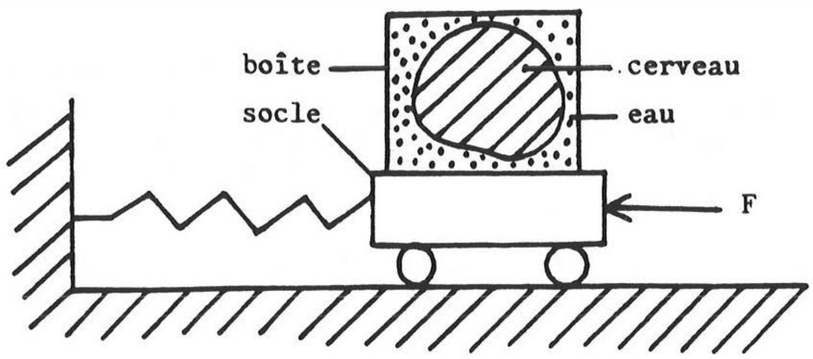
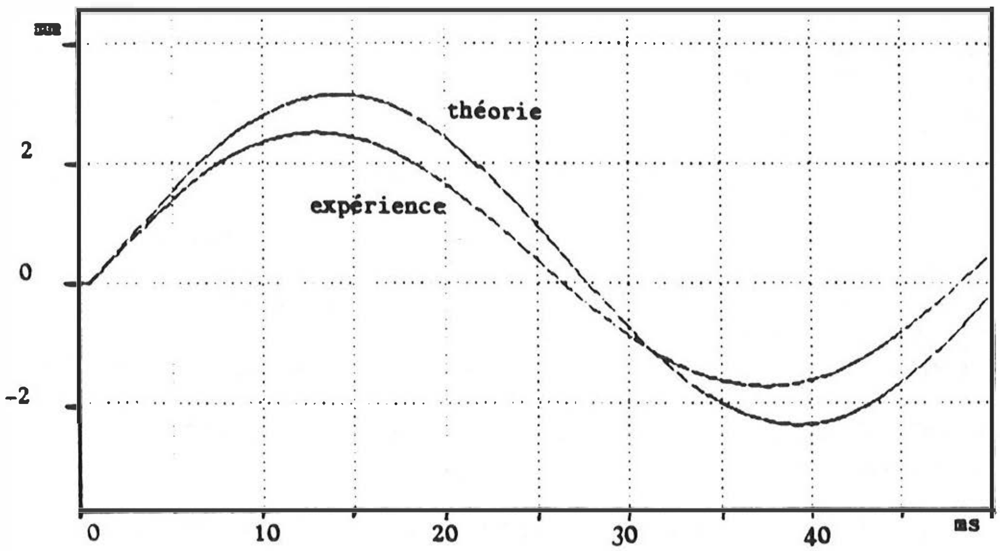


Fig.3: flow diagram of acceleration of the brain through inertial loading

(A)



(B)

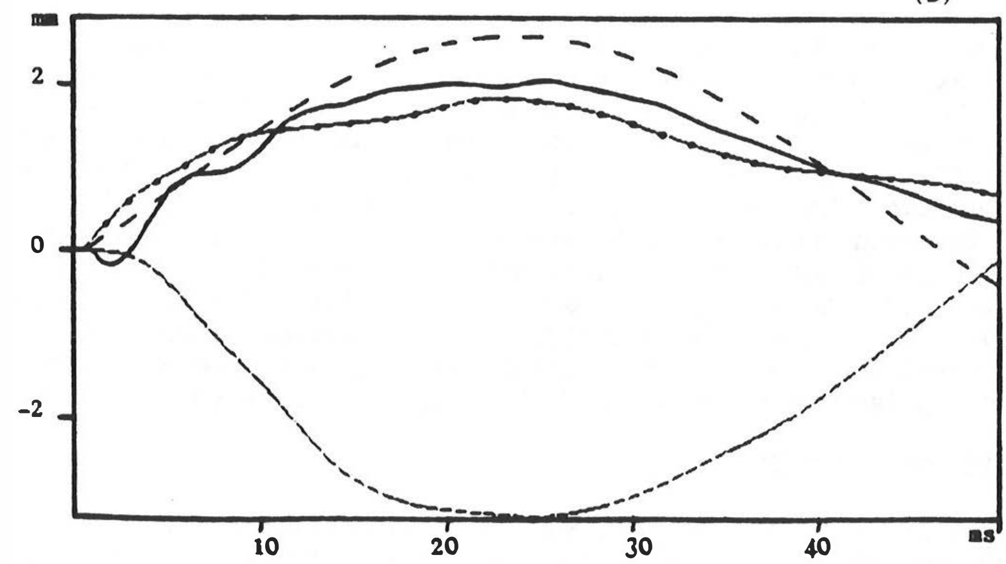


Fig. 4 : Superimposition of the theoretical and experimental displacement A: infinitely rigid base (m= .67 kg)
B: base, box, brain together (m = 1.5 kg)
— — —Theory —•—•—•support ----.brain

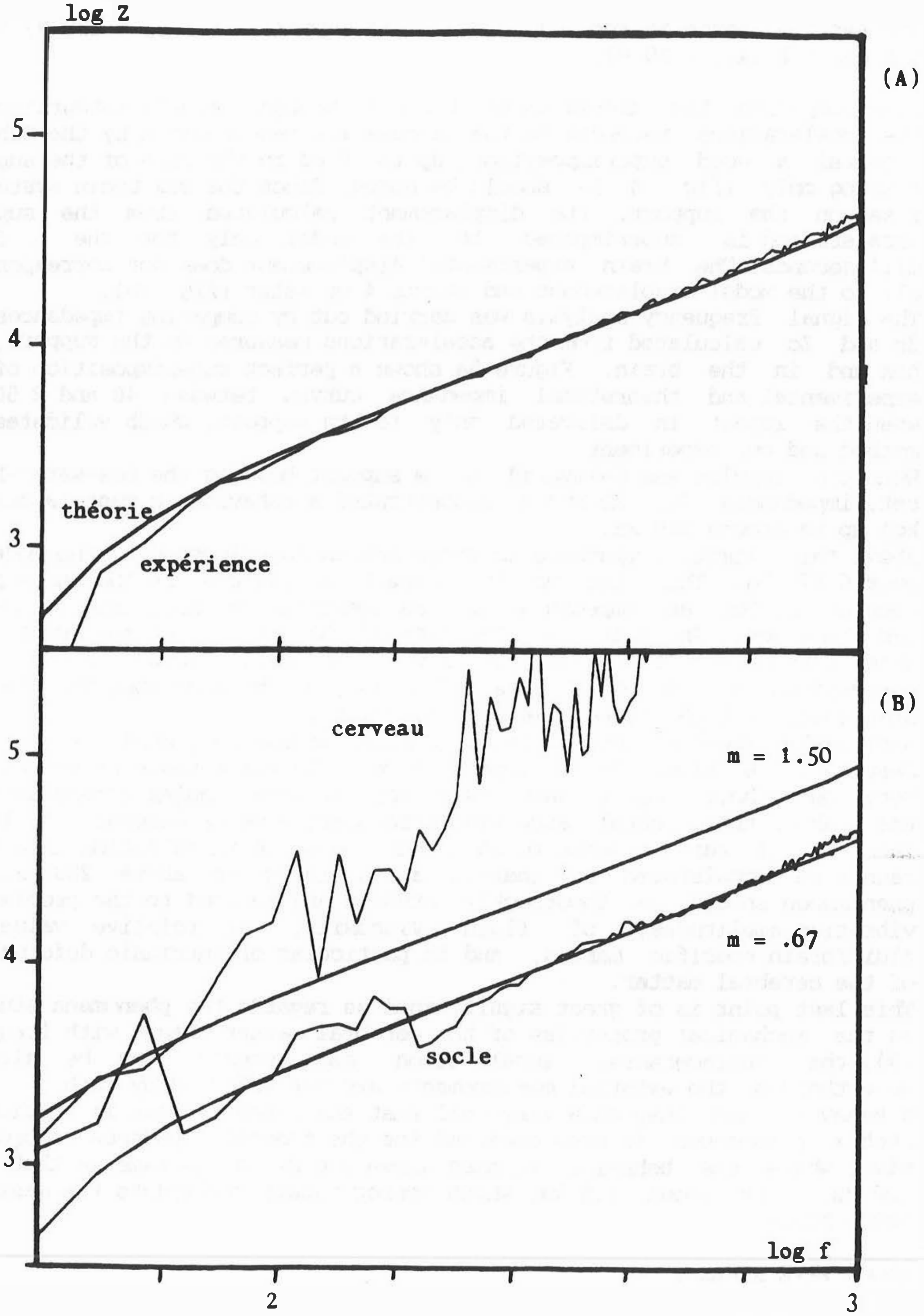


Fig. 5 : Mechanical impedance relative to the test illustrated figure 3
A) infinitely rigid support only (m = .67 kg)
B) support, box, brain together (m = 1.5 kg)

whether the force is applied to the rigid support ($F_{max.}$ = 180 N) or to the box ($F_{max.}$ = 10 N).

When comparing the displacement obtained through a double integration of the accelerations recorded to the calculation result given by the Duhamel integral a good superimposition, up to 10 ms in the case of the support testing only (fig. 4 A), should be noted. Since the box-brain system is fixed on the support, the displacement calculated from the support acceleration is superimposed to the model only for the 5 first milliseconds. The brain experimental displacement does not correspond at all to the model displacement and starts 4 ms later (fig. 4b).

The signal frequency analysis was carried out by comparing impedances Zs, Zb and Zc calculated from the accelerations measured on the support, the box and in the brain. Figure 5a shows a perfect superimposition of the experimental and theoretical impedance curves between 40 and 2 500 Hz when the impact is delivered only to the support, which validates the method and our experiment.

When the impulse was delivered to the support holding the box-water-brain set, impedances Zs, Zb and Zc demonstrated a behavior in mass (about 1.5 kg) up to around 200 Hz.

Above this limit, impedance Zs drops and is superimposed to the model of mass 0.67 kg. This led us to suspect a failure of the box-support connection, but as impedance Zb has identical feature, this assumption was dismissed. In fact, as the mass of the box and the few quantity of water surrounding the brain is low, the mass "loss" above 200 Hz corresponds to the brain mass (0.8 kg), which means that the brain is stationary at high frequencies (see figure 5b).

During the impact at the box level, similar impedance graphs are observed, despite a 20 times lower impact force. The first tests of sudden head rotation without impact show that for variable angular accelerations , the peak of the cerebral tangential acceleration never exceeds 5 to 10 g.

This set of results seems to show that the brain in retention in a fluid cannot be accelerated by inertia above 10 g or above 200 Hz. This phenomenon should be thouroughly studied and related to the problems of vibration amplitudes, of fluid viscosity, of relative values of fluid/brain specific masses, and in particular of intrinsic deformations of the cerebral matter.

This last point is of great significance as regards the phenomena studied. As the mechanical properties of the cerebral matter change with frequency (9), the intracerebral acceleration measurements can be altered. Nevertheless, the external measurements are not affected by such a behavior and they show very well that the cerebral mass is stationary.

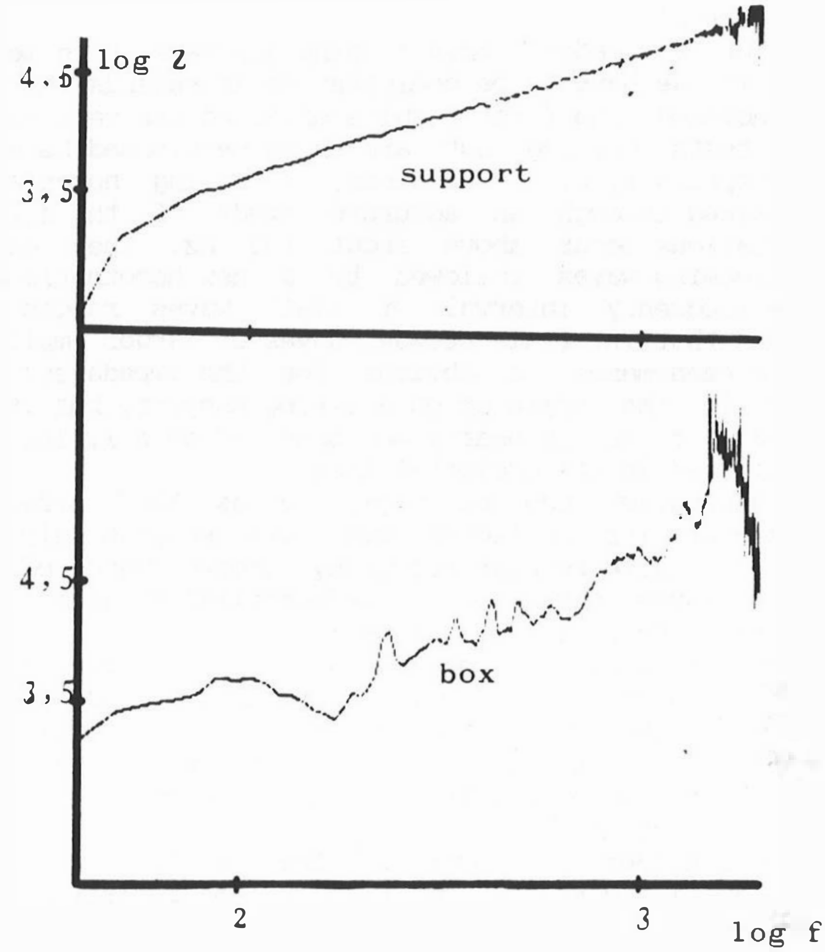
Such a phenomenon is also observed for the frontal impedance recorded in vivo, where the behavior in mass above 200 Hz is the same as that below 100 Hz, less about 1.5 kg, which approximately represents the mass of a human brain.

SHOCK WAVE REFLECTION

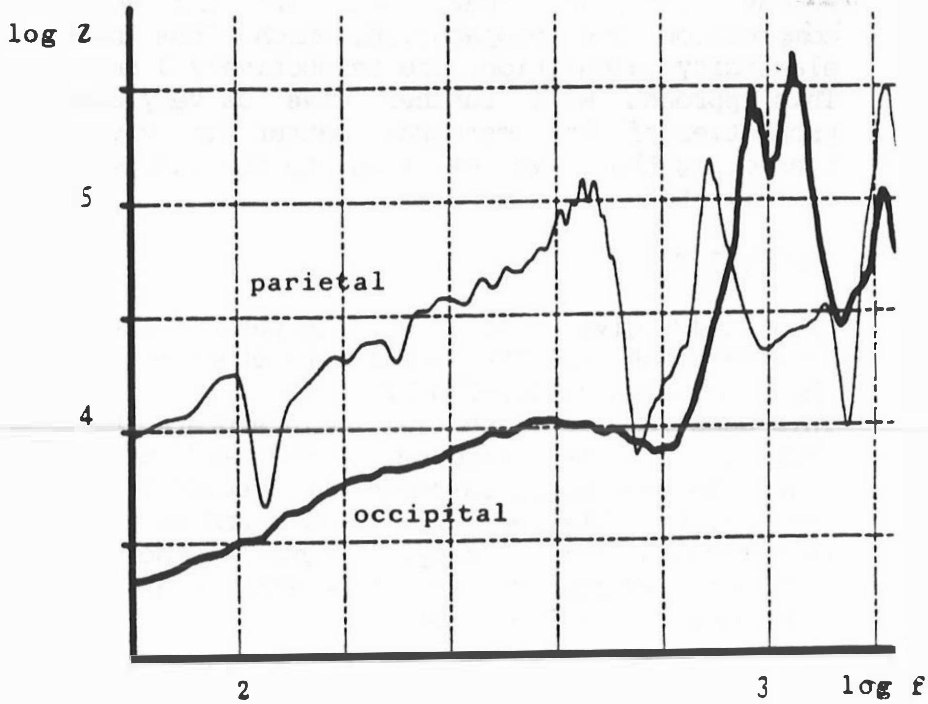
The determination of information data relating to the shock wave reflection in mechanical impedance seemed to be required as the curves obtained were very similar to the hydraulic impedance curves given in the bibliography on blood flows (10). Indeed, if for an infinitely rigid

Fig. 6 : shock wave propagation through mechanical impedance undulation

A. Recordings relative to the test illustrated in figure 3.



B. In vivo recordings with parietal and occipital impact.



system the impedance curve recorded is perfectly smooth, the curve obtained from an accelerometer fastened on a flexible box filled with a fluid or on a skullcase shows regular vibrations around an average values (figure 6).
To our knowledge, this finding has never been mentioned in bibliography, but if we have to be conservative as regards the interpretation of such a phenomenon ,the first results obtained are very encouraging.
The tests carried out are those mentionned hereabove (physical model of box-spring type, 3 cadavers, 1 living human subject). The results are obtained through an accurate study of the different impedance curves. Vibrations occur above about 100 Hz. They often are in the form of 3 decreasing waves followed by a new homothetic series to the first one. The frequency interval of small waves ranges from 30 to 35 Hz, which means that intervals between waves of larger amplitude equal about 100 Hz. This phenomenon is obvious for the impedances recorded on the flexible box, in the brain or on a living subject, but it is hardly determined on a cadaver and is nearly not observed on a living subject when the blow is delivered in the occipital area.
This behavior can be explained as the effects of the shock wave and those of its reflected wave are successively cumulated and equated to zero to give regular frequency interval waves of the hydraulic impedance: these waves correspond to deformations or displacements when acceleration is measured on a flexible wall.
The fact that this phenomenon is not observed on a cadaver or on a living subject when the blow is delivered in the occipital area could therefore be explained by a modification of the brain-skull interface on the cadaver (a detachment is observed) and by the thickness of the skullcase which is twice thicker in the occipital area than in the frontal or parietal areas.
When considering waves of the 35 Hz "period" and an intracranial propagation distance of 0.15 m, the reflected wave propagation velocity is about 7m/s. The waves occuring at 90-100 Hz intervals yield a propagation velocity of the order of 20m/s.
We may consider that these are the velocities of the shear and compression vave propagation, which, from the transverse and longitudinal elasticity calculation, are respectively 3 and 30 m/s.
This approach will further give us very usefull data on the mechanical properties of the cerebral matter in vivo. Investigations aiming at connecting the wave amplitude to the injury parameter of a blow could be contemplated.

CONCLUSION

This study gives the basic answers relating to the influence of the cerebral mass on the modal parameters of the first natural frequency of the human head at 100-200 Hz.
This question is essential to interprete the mass-spring models designed from the impulse response of the head, and then to analyse the injuries using the same modal parameters as lesion indicators.
The results obtained from in vivo and in vitro heads, in addition to the test results for a simple physical model showed that the brain follows the "box" motion up to a frequency of about 200 Hz, and that above this frequency, it is stationary.

This study should be furthered by analyzing the exact causes of such a phenomenon in order to determine whether it is due to a modification of the mechanical properties of the cerebral mass or to the difference, indeed not significant, between the volume masses of the brain and the cephalorachidian fluid.
To end with, the basic data of a measurement technique for the brain mechanical properties based on the recording of the hydraulic impedance of the head are given. In the future, this measurement could be considered as a injury parameter occuring in the cerebral injuries.

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