METHODS FOR BACKING-UP THE CONCLUSIONS OF ACCIDENT RECONSTRUCTIONS CARRIED OUT WITH INSTRUMENTED CADAVERS.-

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. INTRODUCTION

Determination of tolerances of the human body during a car accident have made use, for a few years, of experimental data obtained from fresh non-embalmed cadavers.

The correct undertaking of experiments and the interpretation of results rapidly gave rise to delicate problems - the subjects available are not identical, as are impact dummies. Obviously biological characteristics of cadavers have, at least, the same scattering as those observed at living people. Furthermore, they are different from the actual victims of accidents. As regards the skeleton, cadavers use to be more fragible than living people.

The question proved to be more acute when results of reconstructions of actual accidents became available and in which the differences of severity of the injuries of the actual victim and of the cadaver used in reconstitution often proved to be considerable with the most severe injuries occuring to the cadaver.

We give details below of a first attempt of an overall method of interpreting, with regard to the thorax, results of simulations of frontal impact with 3-point belts, and to explain the differences observed.

This method makes use of results of a simplified mathematic model of the thorax of which the theory is explained after a brief description of the experiments.

In the event of reconstruction of an actual accident, it is important that the differences between the cadavers and the simulated victims be reduced to a minimum. This requires thorough knowledge of the actual accident which is obtained by a multi-disciplinary enquiry: age, anthropometric features of the victim, positions in the vehicle, etc. The kinematic differences may be reduced by selecting the subject and its position.

Furthermore, in any simulation of frontal impact, reestablishment of the volume of the chest by blowing in air by means of a tracheotomy orifice (1) provides a means of obtaining more realistic positions of the diaphragm and of the ribs.

We still have no certain information, after this procedure, on the condition of the bones of the subject before the crash, and work similar to that of Calspan (2) should be effected to fill the gap and define testing to select convenient subjects before the crash.

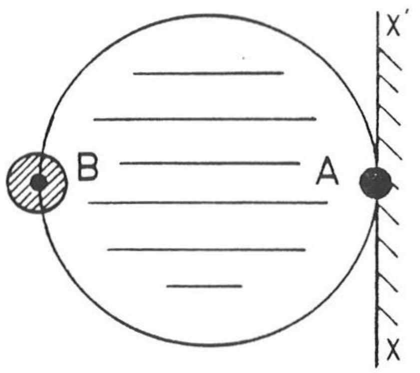
However, the strength of the rib cage which is the part of the body of which the strength is critical in the case of frontal impact with seat belts, may be assessed after the test. Simple mechanical tests of strength of ribs (3) and determination of their mineralisation would supply quantitative data. In this first step, this is purely geometric data - the plane cross-section of ribs which have been used here while waiting for the other available results to be included in the evaluation of strength of the rib cage.

Measurements taken from cadavers are very often measurements of acceleration and of retention forces. Measurements of acceleration are the subject of discussions and research in order to improve knowledge of their relations with the injuries observed (4). We possess measurements of linear accelerations taken at the fourth vertebrae and have used this data without presuming that better measurements of acceleration may be envisaged and utilised later.

Description of the simplified model of the thorax.-

The cinematics of the thorax of an individual wearing a seat belt during frontal impact is well known. The thorax is projected forward but the movement is limited by the seat belt. The internal organs are also projected forwards but they thrust against the thorax wall because the thorax is retained. One of the bases of our modelisation will be to assume that the behaviour of viscera is similar to that of an incompressible fluid which creates a pressure of hydrostatic type inside the thorax cavity; the other basis is modelisation of the skeleton.

The modelisation used is shown in Figure 1.



Ball joints in A and B

Added mass μ in B

Safety belt simulated by a plane (x', x)

Incompressible fluid in thorax cavity.

It is difficult to model the thorax mechanically. We have made an approach which is different to those already effected. It is neither a model or finite elements nor a model based on springs and shock absorbers (5). Rather than attempt to rigorously model the thorax, we have tried to quantify the relative influence of the various parameters, specific to each subject on their state of resistance.

Assumptions - As we have said, the viscera are simulated by a perfect and incompressible fluid in a container, the thorax cavity. This container is not easily modelisable because of its particular geometry. We have idealised its characteristics by assuming it to be cylindrical with a circular base. Furthermore, as a first approximation, we have neglected the effects caused by the height of the thorax and we have considered the problem as a plane, assuming the cylinder to be of infinite length. This means that the study is carried out in two dimensions, leaving aside the effects due to the third. Following the same idea, the safety belt is simulated by a rigid, indeformable plane on which the thorax is thrust.

The stresses which effect the experimental subjects during an impact are complex and vary with time. The thorax itself is the centre of a set of forces and accelerations difficult to assess. The model reveals forces of inertia and we consider that thorax acceleration gives better information on the intensity of the impact, within the scope of the model. Acceleration of the subject will be considered as uniform which leads us to a problem of statics. In order to provide a link with the results of true-scale experiments on cadavers, we made use of data which is available after tests, the 3 ms resultant of accelerations measured at D4 during impact. The excessively transient acceleration peaks are thus mitigated.

Ribs, which compose the cylinder, are assumed to be without weight. The only forces of inertia to be taken into account are those due to fluid and an overload to simulate the backbone and the components solid with it. The laws of behaviour of bones are idealised by neglecting plasticity and visco-elasticity of the material.

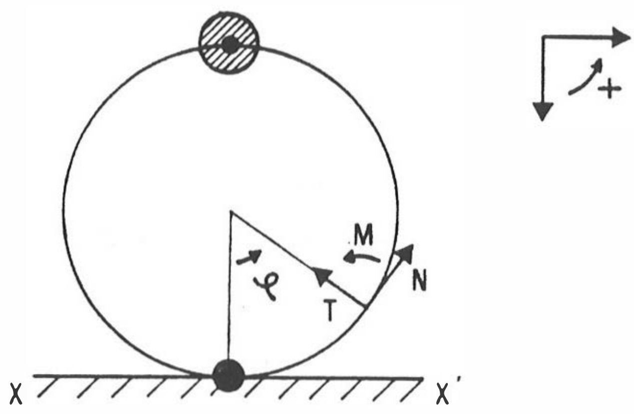
If equations are drawn up for the model as defined above, important figures are found for stresses at the sternum and the rib cartileges; yet fractures are rarely observed there. This may be explained for the sternum which is a thick and resistant bone but not for the costal cartileges. This has led us to make a supplementary assumption.

The behaviour of costal cartileges is different from that of ribs. A possibility exists of movement in relation to costo-sternal connections. Similarly, the costo-vertebral connections are sufficiently flexible to allow limited rotation. We have therefore added ball joints at the sternum and at the back-bone to simulate these rotations.

Mechanical Analysis

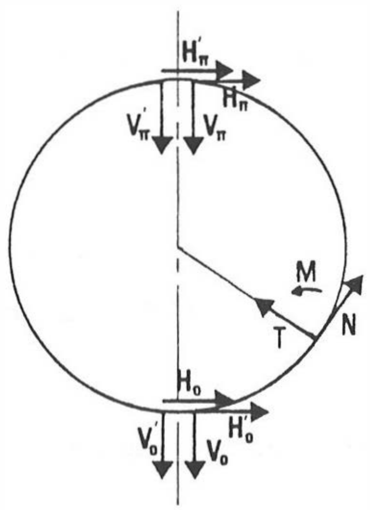
Notations -
- R average radius of the thorax
- φ angle related to a cross-section
- μ added mass, attached to the spine
- ρ fluid mass, per area unit
- γ fluid acceleration
- $M(\varphi)$ bending moment, associated to angle φ.
- $N(\varphi)$ normal force, associated to angle φ.
- $T(\varphi)$ shearing force, associated to angle φ.
- $2V$ reaction exerted by the supporting plane

Hypothesis of resistance of materials were applied. Problem is solved in two steps. The first step corresponds to the action of fluid alone; the second to the action of the added mass μ alone. The two actions are then added.



a) <u>isolated fluid</u>.

According to the symetry of the problem, a cutting is supposed to be made along the vertical axis (O,π). Equilibrium is re-established by introducing the unknown values of M, N and T on points O and π. The equilibrium of the right half is then taken apart.



Because ball-joints are placed in O and :

$$M(0) = M(\pi) = 0$$

The only remaining unknown quantities are:

(H: Horizontal Component, V, Vertical Component of the forces acting in O or to ensure the equilibrium. The stressmark accompanying any letter is related to the left side).

Symetry gives: $V_\pi = V'_\pi = 0$

$V_0 = V'_0 = V$

Besides, three relations describing the equilibrium may be written:

- horizontally: $H_0 + H_\pi + \int_0^\pi p\gamma R^2 (1+\cos\varphi)\sin\varphi \, d\varphi = 0$

- vertically: $V + p\frac{\pi\gamma R^2}{2} = 0$

- equilibrium of bending moments (here $\varphi = 0$):

$$-2RH_\pi - \int_0^\pi p\gamma R^3 (1+\cos\varphi)\sin\varphi \, d\varphi = 0$$

The three relations allow us to obtain the following values:

$$H_0 = -p\gamma R^2 \qquad H_\pi = -p\gamma R^2 \qquad V = -p\frac{\pi\gamma R^2}{2}$$

It becomes then possible to define the required quantities for the cross-section :

$$M(\varphi) = -\left(H_0 R(1-\cos\varphi) + VR\sin\varphi + \int_0^\varphi p\gamma R^3 (1+\cos\psi)\sin(\varphi-\psi)\,d\psi\right)$$

$$N(\varphi) = -(H_0\cos\varphi - V\sin\varphi)$$

$$T(\varphi) = H_0\sin\varphi + V\cos\varphi + \int_0^\varphi p\gamma R^2(1+\cos\psi)\cos(\varphi-\psi)\,d\psi$$

That gives:

$$M = -\frac{p\gamma R^3}{2}(\varphi-\pi)\sin\varphi$$

$$T = \frac{p\gamma R^2}{2}(\sin\varphi + (\varphi-\pi)\cos\varphi)$$

$$N = \frac{p\gamma R^2}{2}(2\cos\varphi - \pi\sin\varphi)$$

b) added mass alone.

The action of the added mass has to be combined to the previously written values.

The method for calculating is the same.

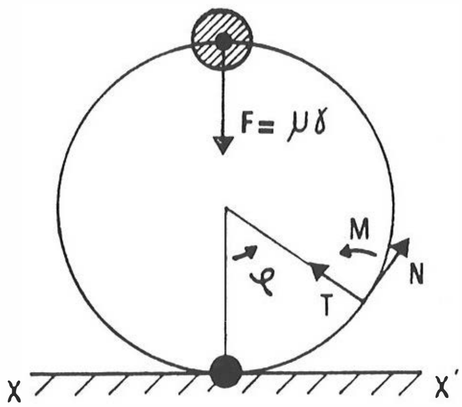
$$M(0) = M(\pi) = 0$$

$$H_0 = H_\pi = 0$$

$$V_0 = V = -\frac{F}{2}$$

That gives:

Figure 4



$$M = \frac{FR}{2} \sin \varphi$$

$$T = -\frac{F}{2} \cos \varphi$$

$$N = -\frac{F}{2} \sin \varphi$$

By combining the effects of the fluid and of the added mass, we have:

- bending moment: $M = -\frac{\rho \gamma R^3}{2} (\varphi - \pi) \sin \varphi + \frac{FR}{2} \sin \varphi$
- normal force: $N = \frac{\rho \gamma R^2}{2} (2 \cos \varphi - \pi \sin \varphi) - \frac{F}{2} \sin \varphi$
- shearing force: $T = \frac{\rho \gamma R^2}{2} (\sin \varphi + (\varphi - \pi) \cos \varphi) - \frac{F}{2} \cos \varphi$

If we make the following hypothesis, i.e. the added mass tied to the spine is equal to 1/5 of the torso section mass, it gives:

- bending moment $M = -\frac{\rho \gamma R^3}{2} (\varphi - \frac{5\pi}{4}) \sin \varphi$
- normal force: $N = \frac{\rho \gamma R^2}{2} (2 \cos \varphi - \frac{5\pi}{4} \sin \varphi)$
- shearing force: $T = \frac{\rho \gamma R^2}{2} (\sin \varphi + (\varphi - \frac{5\pi}{4}) \cos \varphi)$

Variations of M, N and T constitute figures 6 and 7.

Determination of stresses in model.-

The model studied is a plane model. Therefore, if forces applied to a rib are to be known, allowance must be made for its cross-section. We therefore have:

$M_c = M(\varphi) . l(\varphi)$

$N_c = N(\varphi) . l(\varphi)$

$T_c = T(\varphi) . l(\varphi)$

in which $l(\varphi)$ is the width of the rib, M, N and T are the forces calculated in the plane model for an angle φ. If it is desired to calculate stresses on a cross-section of a rib, it must be assumed that the rib in question is located in a horizontal plane. The forces on a cross-section marked by the angle φ may be calculated on this condition.

Normal stress:

$$\sigma(\varphi) = \frac{N_c(\varphi)}{s(\varphi)} + \frac{M_c(\varphi)}{I(\varphi)} y(\varphi)$$

Tangential stress:

$$\tau(\varphi) \simeq \frac{T_c(\varphi)}{s(\varphi)}$$

When $M_c(\varphi)$ = bending moment in φ

$N_c(\varphi)$ = normal force in φ

$T_c(\varphi)$ = shearing force in φ

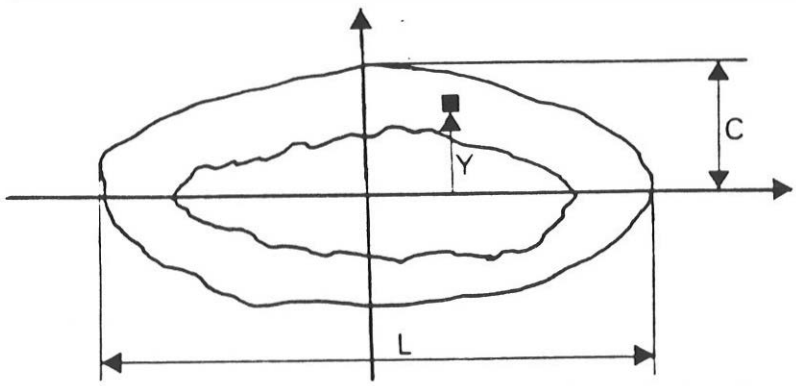
$s(\varphi)$ = area of cortical bone in φ)

$y(\varphi)$ = distance from point considered to neutral axis in φ) see figure

$I(\varphi)$ = inertia of section in φ)

With $\sigma(\varphi)$ a maximum for y maximum, it is later considered that $y = C$ when C is the maximum distance from the neutral axis. Maximum stress is thus calculated.

Figure 5



We do not possess a list of cross-sections for the whole thorax circumference, for each subject, but only one or two cross-sections taken from the anterior/posterior arch of the 4th, 5th and 6th ribs. We can therefore only calculate forces for one or two values of $C(\varphi)$, $I(\varphi)$, $s(\varphi)$, $\ell(\varphi)$, without exactly knowing the value of φ related to the cross-section. However, the cross-sections studied were all cut from similar positions and it may be assumed that the values of $\frac{\ell c}{I}$ and ℓ/s, geometrical ratios which are used in calculation of forces, are representative, overall, of the ribs of the subject.

Calculations show, furthermore, that $\frac{Nc}{s}$ is negligible when compared with $\frac{M_c C}{I}$.

Stresses calculated become:

- Normal stress: $= \frac{M_c C}{I}$

- Tangential stress: $= \frac{Tc}{s}$

Maximum stresses are then calculated for each subject. The average radius of the thorax taken for calculation is approximated by: R = Thorax perimeter/2π

$$\sigma(70°) = 2.5 \frac{\rho \gamma R^3}{2} \frac{\ell c}{I}$$

$$\tau(0°) = 3.9 \frac{\rho \gamma R^2}{2} \frac{\ell}{s}$$

ρ is taken as constant for all subjects and equal to 1000 kg/m^2.

Comparison of subjects - The number of rib fractures is noted on autopsy for each case of accident simulation using a cadaver for which all necessary data are available. The corresponding points are plotted on graphs, number of fractures against normal stress and number of fractures against tangential stress.

Utilisation of results - The dispersal of results shown in graph (NF, τ) is wide, particularly in view of the small number of cases studied. The method does not therefore seem to be very suitable for showing the response of the thorax to possible shearing phenomena.

On the other hand, a certain trend is shown in graph (NF, σ). In fact, if two aberrant points are eliminated (which we shall explain), it would seem that the number of fractures observed is a linear function of stress calculated in the model when $\sigma > 40$. A phenomenon of threshold is seen when $\sigma < 40$; it seems that 0, 1 or 2 fractures may be found in an individual in the same conditions of slightly violent impact. This graph is of interest for it enables, if confirmed by a greater number of experiments, a forecast to be made of the number of fractures found in a body subject to experiment and of which the characteristics are known, to within one or two fractures, or at least, to establish significant areas. One might propose as an example:

$\sigma < 40$ daN/mm^2	$0 \leq NF \leq 5$
$50 < \sigma < 150$ daN/mm^2	$7 \leq NF \leq 12$
$\sigma > 150$ daN/mm^2	$NF \geq 12$

An observation is necessary at this stage. The values for stresses obtained do not rigorously represent those which might be observed in reality, but serve to define a scale of comparison. Following the same idea, acceleration of the thorax used in the calculations cannot be compared to that measured on dummy.

This method would be of no use if it could not be applied to forecasting injuries incurred in a true accident; this is the purpose of the following.

Extrapolation of the method to living persons - This method is not directly applicable to people actually exposed to the risk. It is not actually possible to have direct knowledge of the condition of bones of living people. The strength of experimental subjects is often less than that of the average population. In order to gain knowledge of the strength of living persons exposed to the risk of accident, samples of ribs are taken from persons who have died suddenly, poisoning, suicide, accident, without a long stay in hospital and their characteristics are studied - cross-section, mineralisation, etc - assuming that they are representative of the living population.

Thus the values of $\frac{\rho_c}{I}$

and $\frac{e}{s}$

may be calculated within the scope of the model for subjects who have died suddenly. It is noticed that these values are lower, on average, than those calculated on cadavers and are, above all, less dispersed. $\frac{\rho_c}{I}$ varies from 0.2 to 0.6 whereas tests on cadavers gave a $\frac{\rho_c}{I}$ varying from 0.2 to 1.2.

Application of method in reconstruction of a fronto-frontal accident - The subject (6) is a cadaver tested in a frontal impact and wearing a belt within the scope of a reconstruction of an actual accident. He simulated the right front passenger. The table below gives a comparison between the two accidents, actual and simulated, with respect to the thorax.

	Real victim	Cadaver
Sex/Age	F/30	F/54
Weight	54 kg	47,5 kg
Height	1,60 m	1,53 m
γ 3 ms	--	50 g
Number of rib fractures	0	10
$\frac{\rho_c}{I}$	--	0,45 mm^{-2}
C/M	--	0,26
σ	--	50,4 daN/mm^2

As the anthropometric features of the actual victim are very similar to that of the cadaver of the experiment, it may be assumed, in order to make use of the model, that the dimensions of their thoraxes are the same. The features of the bones of the victims cannot be known and so a value is taken for $\frac{\rho_c}{I}$

equal to that of the average of people autopsied after sudden death (0,36). is then calculated as 40 daN/mm^2 that is to say, at the threshold of appearance of fractures which have been found (i.e. 40 daN/mm^2). Near to the threshold, there is an important lack of precision in forecasting the number of fractures. In order to explain the relatively favourable results that occurred to the actual person involved in the accidents, use has been made of mineralisation of ribs. The ratio of weight of ash to the initial weight of the fresh ribs was 26 %, but the ratio would be of the order of 35 % for persons who have died suddenly. The mineralisation factor can only have an effect with respect to increase of strength of ribs and may suffice to explain that the contemplated passenger was undamaged after the accident, at the thorax.

DISCUSSION

a) Subjects not taken into account in the analysis.-

Behaviour of the subjects 7 and 11 is considerably different from that of other subjects on which experiments were made. This is explained by two facts. No 11 is a driver wearing a seat belt, and who was subjected to violent impact of the thorax on the steering wheel. The method of application of forces is different from that considered in the other cases for the steering wheel created a small-size overload which can be allowed for in the model. No 7 was, on the other hand, very slightly injured in comparison with what might have been forecast from the model. This has been explained by the greater degree of mineralisation than that of the average of subjects on which experiments were carried out. (0.35 instead of 0.27).

No 7 was accordingly not taken into account in the present state of the study - temporarily -.

b) Validity of the model.-

The sample in question only concerns subjects wearing seat belts who have been subject to a frontal impact at speeds of about 50 km/h. It would seem, after a few tests at 65 km/h, that the response of the model is no longer suitable and does not allow comparison between subjects which have been subject to impacts at very different speeds. This particularity is doubtless due to faulty comparison between the violence of the impact and the particular thorax acceleration which we have used. A next stage will consist of determining what acceleration parameter (or connected to it) best relates to acceleration of the thorax.

The model as it is does not take account of the mode of application of forces on the thorax. The area of contact with the seat belt in particular is not taken into account.

c) Improvements.-

The only parameter selected which characterises the ribs of subjects was $\frac{\ell_c}{I}$, geometrical data of a rib cross-section. This factor appeared in calculation of rib strength but it is not sufficient to characterise the overall strength of subjects. We have assumed that $\frac{\ell_c}{I}$ is sufficient basing ourselves on the fact that subjects become osteoporotic with age but, in general, with no osteomalacia (3), that is to say that the area of cortical bone diminishes without damage to the quality of the bone. This affirmation should however be moderated and, in a second stage, mineralisation of subjects should be allowed for.

CONCLUSIONS

By using the results of a small number of simulations of frontal impacts, the small number being due to the necessity for having certain data available simultaneously, it was possible to link the number of rib fractures observed with the anthropometric features of a cadaver, the geometry of his ribs and a parameter which depended on acceleration of the thorax.

The conclusions must be validated by a greater number of tests and refined by improvement of the model itself and by an increase in the number of parameters considered by, for example, making use of mineralisation.

The study carried out in its present imperfect state provides, nevertheless, a comparison of the great differences of severity of injuries which are seen in the same accident from one cadaver to another, or between a cadaver and the actual victims of the accident.

This work is the first stage of research on evaluation of tolerance of the population exposed to the risk of accidents and of a comparison of systems of retention.

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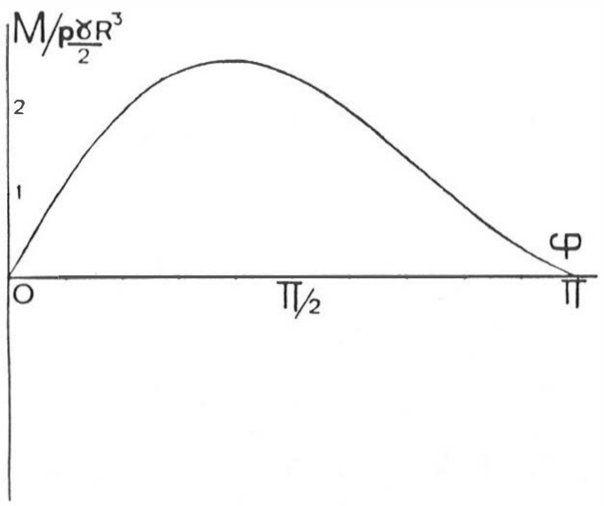
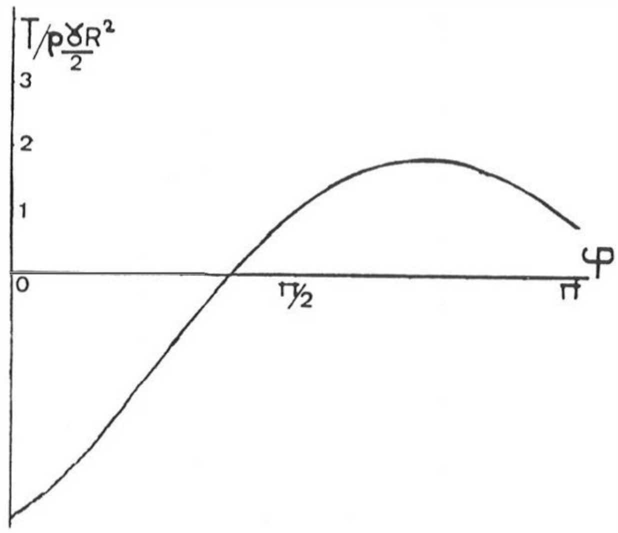


Figure 6 - Bending moment

Figure 7 - Shearing force



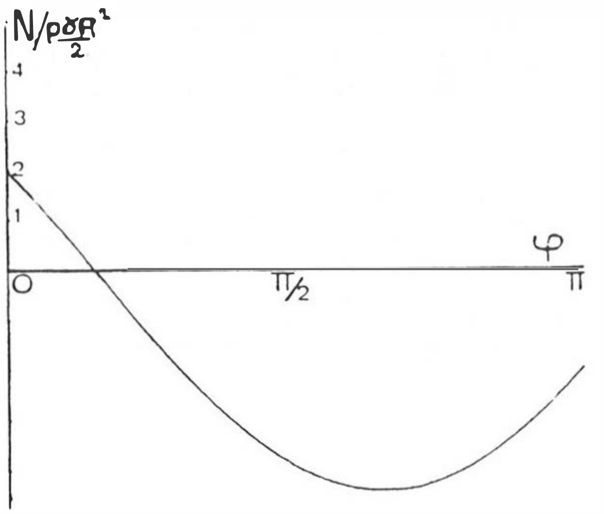


Figure 8 - Normal force

Cadavers	Sex	Age	Height	Weight	Thoracic Perimeter	Thoracic Widthness	Thoracic Thickness	Rib characteristics						Number of rib fractures	σ daN/mm²	ε daN/mm²	Comments
								L	C	I	S	C/M	γ3ms (g)				
1	M	51	171	50	795	275	180	10,7	3,9	148	23,6	0,27	40	2	29,5	0,75	
2	M	57	164	68	865	285	220	11,1	2,67	28,2	7,6	0,23	23	9	78,8	1,24	
3	F	59	158	60	857	270	195	8,33	2,47	18,4	5,3	0,18	37	10	131,2	2,1	
4	M	53	174	66	880	260	210	12	3	30,3	7,1	0,19	50	13	204	3,2	
5	M	64	171	63	898	280	220	10,7	2,67	24,7	7,1	0,22	50	15	211	3	
6	F	54	153	47,5	765	230	190	9,8	3,8	83,4	14,4	0,26	50	10	50,4	0,98	
7		55	155	42	780	250	140	12,1	2,53	41,7	15,9	0,35	47,5	1	83	1,08	
8		52	172	63	925	300	220	14	4,33	232	29,3	0,26	46	7	48	0,92	
9	M	60	163	53	877	270	210	11,3	4	137	21,8	0,35	62	10	70	1,22	
10		52	170	76	960	310	220	9,2	2,4	45,6	22,6	--	50	10	108	0,92	
11	M	51	169	67	950	290	185	14,3	4,27	261	29,6	0,27	100	17	101	2,15	impact/steering wheel
12	M	55	180	96	1130	340	200	10,8	4	75	15,8	0,29	52,5	14	219	2,26	
13	M	58	172	98	1020	310	250	10,7	3,33	103	23,3	--	51	8	94,8	1,2	

Tableau - Data used in the study

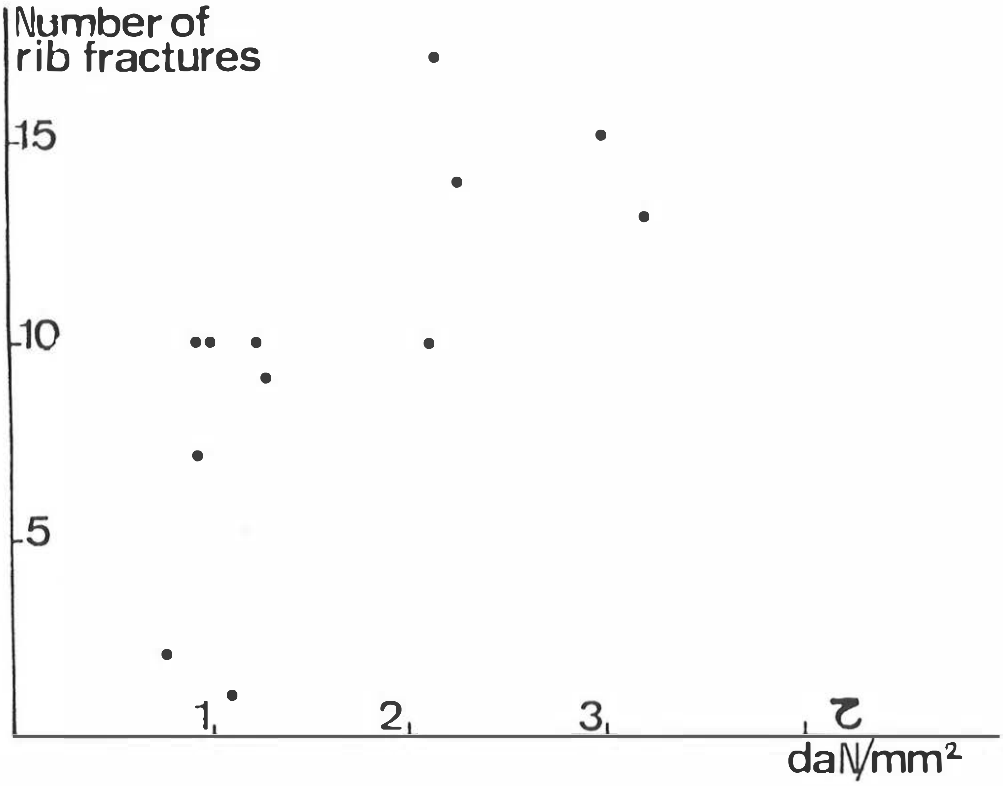


Figure 9 - Graph (NF, τ)

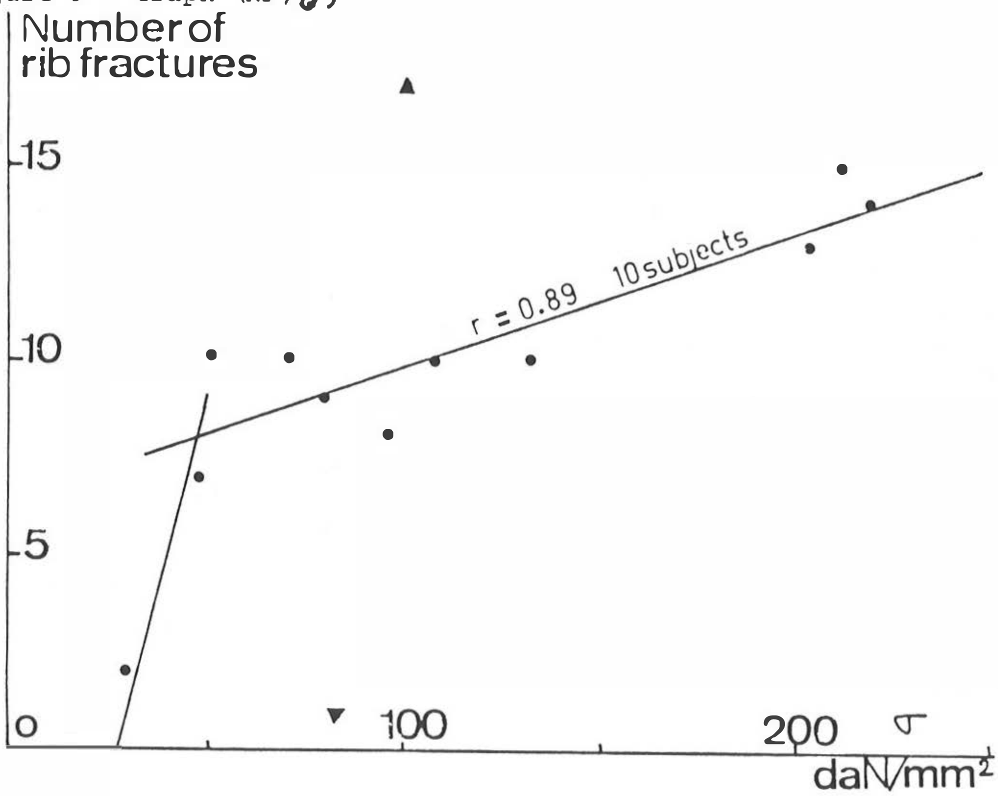


Figure 10 - Graph (NF, σ)