DETERMINATION OF KNEE-FEMUR-PELVIS TOLERANCE FROM THE SIMULATION OF CAR FRONTAL IMPACTS

by

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INTRODUCTION

Many authors have proposed different tolerance criteria for the femur based on differing test methodology (1 to 13). We can observe major discrepancies between these criteria which range from 4 kN to 17 kN. This is partly due to the diversity of test conditions having an effect on results. As an example, we can point out:

- knee impacts against a stop piece, with the subject positioned on a decelerated sled (1 to 4)
- direct impact against the knee by various processes (pendulum impactors or free falls) (5 to 12)
- reproduction of knee impact markings as found in actual accidents (13)

Theoretical analyses were also performed (14, 15).

In the different research programmes, the human subject was never seated in a vehicle, in a frontal impact situation and the experiments were not, therefore, sufficiently representative.

Furthermore, as the load sustainable by the femur is related to the total duration of stress to the knee, and hence closely linked up with test conditions, this duration must be considered as an important parameter in determining the knee-femur-pelvis tolerance. To determine a criterion which can be practically applied in conditions approaching reality, tests were conducted in an "automobile environment" with the occupant subjected to a series of loads comparable to those a real-life victim would experience. The dummy (or human subject) is installed, belted into a standard production vehicle body mounted onto a decelerated sled. Instruments to measure the axial load on the knee were set up and installed in the vehicle body in front of the knees of the test subject. The instruments were fastened directly to the vehicle body and covered with shock-absorbing material. Resultant knee impact durations are in general greater than those submitted in publications by other authors but they are closer to what would have happened in reality in a road situation. To confirm this assumption results obtained from reconstructions of actual accidents are given together with accidentology data. The results of a series of tests conducted with human subjects do not invalidate the highest tolerance levels for the knee-femur-pelvis area, submitted so far.

ACCIDENTOLOGICAL DATA

IRO/PEUGEOT S.A./Renault accident files contain 902 restrained drivers and 470 restrained right front passengers, involved in frontal impact.

Out of 86 seriously or fatally injured drivers, 28 (32%) sustained femoral fractures, and 12 (23%) right front seat passengers out of 51. OSI distribution of these 40 occupants indicates a death rate which is seven times higher than the one observed for the overall restrained occupants in frontal impacts. Indeed, these occupants sustained violent impacts; in 38 % of the cases, ΔV values are greater than 55 km/h with important intrusions because, in 52 % of the cases, the dashboard backward movement on the occupant is superior to 25 cm (table 1).

Table 1 - Dashboard backward displacement

	<15 cm	16-25	26-35	36-45	46-55	>55 cm	Total
Drivers	3	9	9	4	2	1	28
Passengers	2	5	1	2	1	1	12

Fractures localizations:

Table 2 - Nature of the femoral injuries

	inferior extremity	shaft	superior extremity	unknown	Total fractures
Drivers	9	16	7	2	34
Passengers	2	9	2	1	14

For about half of the cases (55 %), fractures are situated at the level of the shaft (average third); fractures of the inferior extremity account for 24 % and of the superior one for 20 %.

Fractures of the two femurs are observed for 2 drivers and 2 passengers (Cf. Table 3 hereafter).

On the other hand, 4 drivers sustained a double femoral fracture, all situated at the level of the shaft and of the superior extremity. This type of injury has not been observed for passengers.

In 10 cases, femoral fractures are associated with bony pelvic fractures; only 50 % of the femoral fractures are situated at the superior extremity, the other 50 % are at the level of the shaft (9 drivers, 1 passenger).

The 11 knee or patella injuries observed in the sample are associated with femoral fractures on the inferior extremity in 7 cases.

Thus the accidentological analysis enables to study the injury frequency of the knee-femur-pelvis area, the nature of these injuries and the violence at which they appear. Each line of table 3 defines an injury association.

Table 3 - Femoral fractures localizations and association with pelvis or knee injuries - 1°) Drivers, 2°) Passengers. N = number of injured members.

1°) 28 Drivers

Knee patella	inferior extremity	shaft	superior extremity	pelvis	N
		X			6
		X	X		2
X		X			1
		X		X	5
		X	X	X	2
	X				3
X	X				6
			X	X	1
X			X	X	1
			X		1-a*
	X	X	X		2-b*
8	9	16	7	9	30

a*: killed non autopsied with unknown pelvis injuries
b*: femoral fracture not precisely located

2°) 12 right front passengers

Knee patella	inferior extremity	shaft	superior extremity	pelvis	N
		X			7
X		X			2
	X				1
X	X				1
			X		1
			X	X	1
	X	X	X		1-a*
3	2	9	2	1	14

TEST METHODOLOGY

Description and installation in the vehicle of knee stop pieces - A system was set up to carry out independent measurements of impact loads against each knee. Two shear sensitive devices, each fitted with a gauge half bridge are used for measurement on each knee. The devices are both mounted in a bracket supporting on either side the round buffer with the knee impacts. The fixture allows a certain amount of independence of measurement in relation to the point of load application. The impact load recorded for each knee corresponds to the normal component of the resultant.

The two stop pieces are fitted to a rigid plate fastened vertically inside a Renault 18 car body, in front of the front passenger's knees, as shown in figure 1.

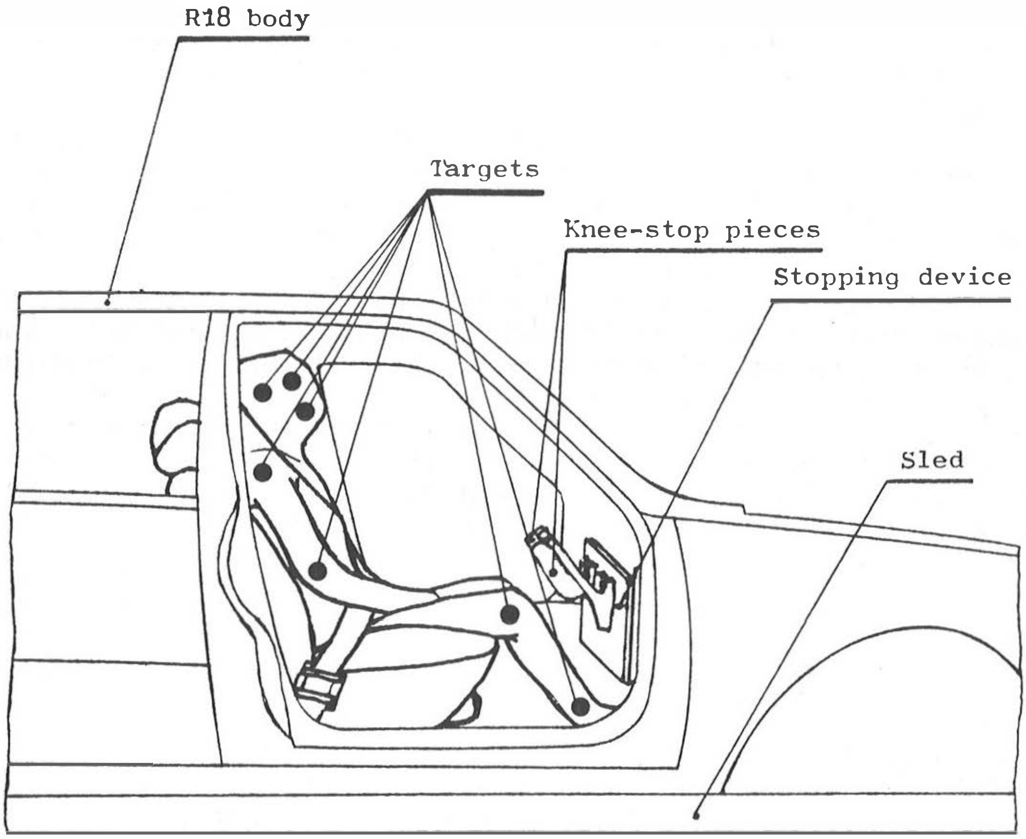


FIG. 1 . Position of the dummy or cadaver inside the vehicle body

Spacing of the two knee stop pieces axes (23 cm) is equivalent to the distance separating the axes of the knees of a 50th percentile such as the Part 572 dummy placed in a vehicle in accordance with the procedure specified in US standard 208. The distance between the knees and the stop pieces may be adjusted by varying the position of the seat. The height of the stop pieces from the floor is adjustable by means of the attachment to the body, to enable correct knee impact, on account of anthropomorphic differences between subjects. The stop pieces must be so oriented that when the knee impacts, the maximum of the impact induced load passes through the femoral axis. A maximum normal component of the resultant must be obtained.

A special study has been conducted to examine this problem. Indeed, knee trajectories are different depending on the vehicle model, the seat used, lap seatbelt restraint... The stop piece buffer angle was determined from a study of restrained passenger kinematics during frontal impacts; the angle is 50° from horizontal.

For test purposes, the buffers are covered with 2.5 cm thick polyurethane foam with a specific gravity equal to 0.115.

Tests and test conditions - The first tests were run at a speed of 50 km/h and a second series at 65 km/h. They were frontal impact tests against a wall.

The vehicle body is a Renault 18 production one. It is mounted on a sled equipped with retarder tubes, allowing sled deceleration to be selected. A number of vehicle components were removed, including the windshield and instrument panel. Front doors were replaced by dummy ones to allow filming. Likewise, an opening was made in the roof, so as to obtain a bird's eye view.

The initial moment of impact is determined by means of a switch which triggers a flash as soon as the sled retarder tubes comes into contact with the wall.

The knee stop pieces are placed, as described above, in front of the front passenger's knees. Just before the test, the foam is coated in white so as to be able to locate knee impact in relation to the centre of the buffers at the end of impact. Foam is added to the lower aperture cross member to limit the violence of a possible head impact.

The dummy or human subject is positioned in the front right hand passenger position. It is restrained by a production three-point inertia type belt, which is changed after each test. The seat features a head-restraint and; seat rake and slide adjustments are carried out at the last minute, to suit the morphology of the test subject. When it is correctly positioned in the passenger compartment, sights are arranged on its different body segments to be able to follow their displacement throughout the impact (fig. 1). All the distances between the targets and, between the knees and stop pieces are measured before impact. An on-board camera on the sled provides a side view in close up.

The following measurements are taken:

- vehicle speed
- vehicle deceleration at "B"-pillar level
- accelerations of thorax
- a three directional acceleration at pelvis level
- knee loads (left and right) recorded on the buffer sensors
- belt loading at different levels (shoulder, buckle, outboard lap, lap buckle).

All the longitudinal components of accelerations are integrated twice.

An initial test to validate the methodology used was carried out with an impact dummy. This was followed by ten or so tests using human subjects. The human subjects used are fresh, unembalmed cadavers which undergo special treatment prior to the test; they are tested less than four days after death, having been preserved in the meantime in a cold room between 0 and 2° C. Their selection was based on bone quality, all those subjects which had been in accidents involving the lower members or which had suffered from bone diseases were not retained for this type of test. Autopsies were then carried out on the cadavers and we were provided with a list of injuries and mention of any fractures in the pelvis-femur-knee region.

Anthropometric data concerning the subjects along with impact conditions to which they were subjected are given in table 4. The results of these tests are given below.

TEST RESULTS

The first test was carried out with a dummy to check instrumentation and compare results with those obtained with cadavers.

Test with dummy - Sled decelerations and standard dummy measurements are recorded in addition to buffer loads. Sled speed was 50 km/h and stop-

Table 4 - Anthropometric data concerning the cadavers and corresponding collision speed

Test No.	Sex	Age	Height (cm)	Weight (cm)	Collision speed (km/h)
231	M	60	165	61	50.7
232	M	57	163	49	50.7
233	M	56	173	63	50.1
254	M	63	162	52	49.5
255	M	68	165	56	50.9
257	F	42	155	53	67.1
258	M	42	164	69	65.5
267	M	68	164	71	60
268	M	62	172	66	66.8
276	M	55	180	82	65

ping distance 600 mm (γ max = 30 g). The knees were positioned to correctly impact the buffers. Knee impact velocity is in the region of 6 m/s along the tangent to the trajectory of the knees when they impact. This velocity cannot be compared to those measured in sled tests presented in the previously mentioned texts, as those tests were carried out with unbelted dummies.

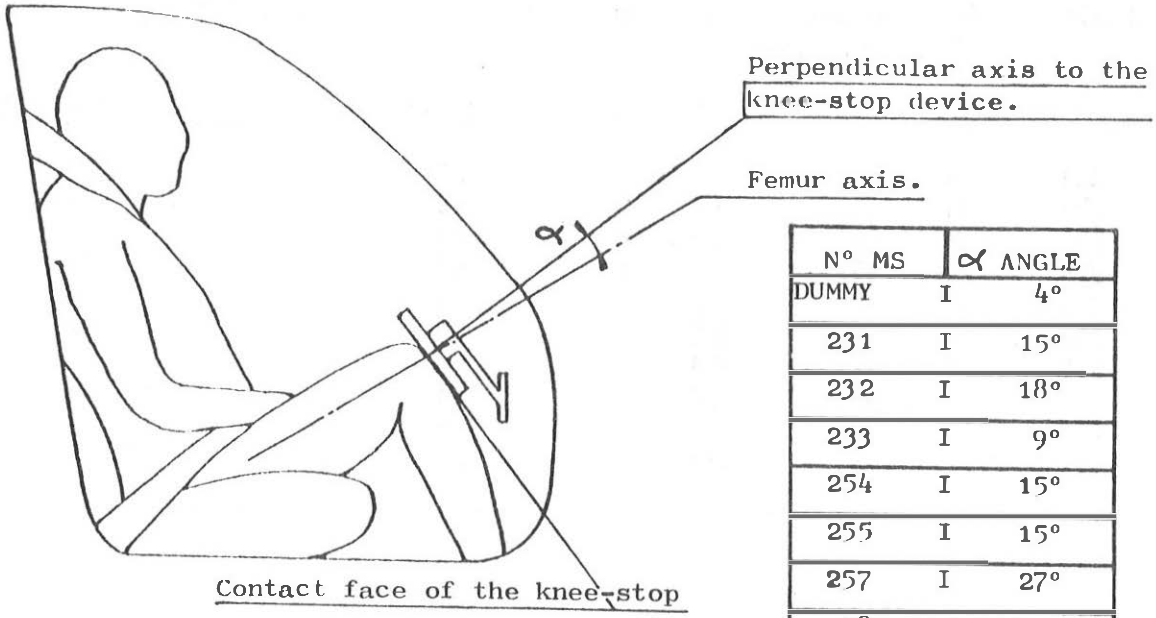
Maximum load values recorded at the knees and femurs are given in the table 5. Mean loads for both knees and both femurs are 16.85 kN and 14.3 kN respectively.

(1) $\frac{\text{left femur}}{\text{time}}$ (kN/ms)	(2) $\frac{\text{right femur}}{\text{time}}$ (kN/ms)	(3) $\frac{\text{left knee}}{\text{time}}$ (kN/ms)	(4) $\frac{\text{right knee}}{\text{time}}$ (kN/ms)	(5) (1)/(3)	(6) (2)/(4)	(7) $\frac{(5)+(6)}{2}$
15.3/11	13.3/10	16.2/11	17.5/10	0.94	0.76	0.85

Table 5 - Results obtained for the dummy test/

Knee impact was analyzed on the films to study in closer detail the femur axial load. This analysis confirms, as expected considering initial test conditions, that when the knees begin to impact the stop pieces, the tangent to their trajectory is approximately perpendicular to the buffer (fig. 2). The difference between the normal load component and the resultant is negligible. The correction was, however, taken into consideration in the results table.

The figure 3 shows the load curves as a function of time, measured at the dummy right and left hand femurs and at the knees stop pieces. The "primary loads" are defined on these curves and the durations of these loads are calculated using the triangular approximation method (3). It is the durations of these primary loads which are taken into account in previous publications



N° MS		α ANGLE
DUMMY	I	4°
231	I	15°
232	I	18°
233	I	9°
254	I	15°
255	I	15°
257	I	27°
258	I	15°
267	I	12°
268	I	6°
276	I	0°

FIG. 2 .Definition of angle α and values of this angle for dummy and cadaver tests.

and not the total durations of load application. The ratios of the maximum femur compression load to the maximum load applied to the knee on the same side are 0.94 and 0.76 for the left and right hand sides respectively, with a mean value for the combined left/right hand of close to 0.85. According to a past study (12) of the response to knee impact of dummies or human subjects sled tests at different speeds, this femur load/knee load ratio is different from the dynamic knee contact load, because of the mass between the femur load cell and the knee joint in the dummy structure.

Tests with human subjects - Ten tests were conducted with unembalmed cadavers. Methodology is identical to that used in the dummy test. There was, however, a change during this series of tests; the fact that there was no fracture of the knee-femur-pelvis during the first tests caused us to heighten the severity of impact by increasing impact speed (65 km/h) and reducing sled stopping distance. The knee impact points are correctly located against the buffers. A study of knee trajectories enabled determination of the direction of the tangent to this trajectory when the knee is contacted and when maximum load is applied. The above figure 2 gives the values of the α angles formed by this tangent with the perpendicular to the buffer for both the knees in each test, when the load is maximum. These values do not exceed 18°, which means,

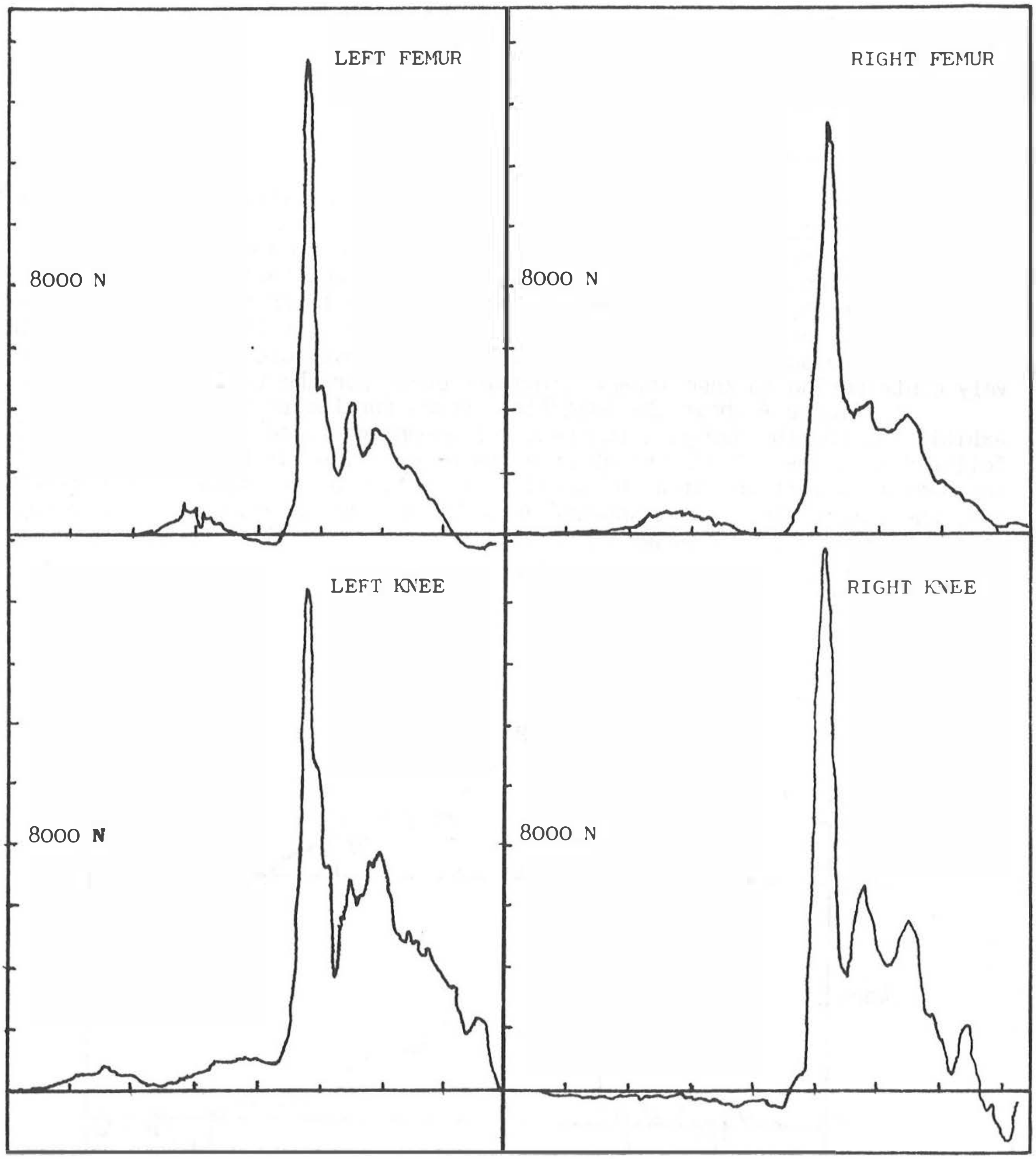


FIG. 3 . KNEE AND FEMUR LOAD CURVES FOR THE DUMMY TEST.

just as for the dummy test, that the tangent to the knee trajectory is almost perpendicular to the buffer. Resultant loads were calculated, taking the values into consideration. Knee orientation in relation to the buffers in the transverse direction was not considered as the angle values were negligible, compared to α.

These resultant loads were normalized to allow for subject mass, using the Eppinger formula (16):

$$\text{Normalized F} = \text{measured F} \times \left[\frac{75}{\text{subject weight}}\right]^{\frac{2}{3}}$$

They are given in Table 6 along with the durations of primary loads as defined above.

An initial remark is called for. Under test conditions strictly identical to those of the dummy test (No. 231, 232 and 233), at a collision speed of 50 km/h, the buffer loads recorded are far lower than those recorded on the dummy. This may be explained for the most part by the difference in the bone structure and flexibility of both types of substitutes, the mass effectively contributing to knee impact being different for the two types of tests.

Figure 4 shows the load/time curves for two tests. These curves exhibit, as for the dummy, a twin peak corresponding to an initial major load followed by a lower load, though of a longer duration. In figure 4, which corresponds to a test in which the patella and pelvis of the subject were fractured, the initial load has a shorter duration and the second peak is less apparent; the knee load decreases steadily.

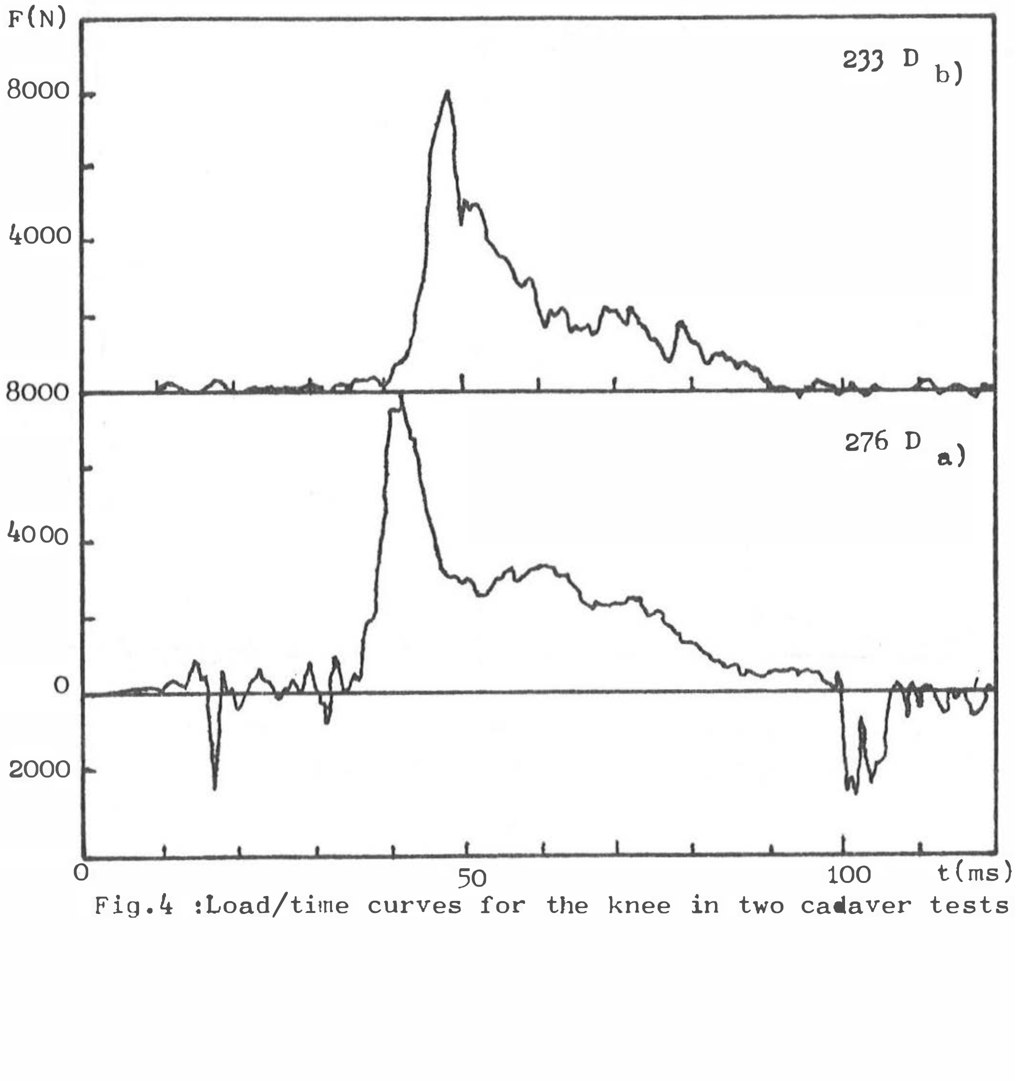


Fig.4 :Load/time curves for the knee in two cadaver tests.

Table 6 - Results for the cadaver tests

Test No.	Peak force/duration right knee - N/ms	Peak force/duration left knee - N/ms	Impulse right knee N.s	Impulse left knee N.s	Fracture
231	7980/11	3920/8	92.4	46.2	
232	10070/18	5030/8	111	43.8	
233	8840/9	9410/18	112.2	99.4	right patella right iliac crest
254	8750/13	9800/17	65.6	65.2	
255	11370/27	9600/18	166	121.4	
257	3670/11	5090/9	18.6	36	
258	8960/13	7580/11	133.2	94	
267	9030/16	6380/14	159.4	79.8	
268	-	6790/17	-	109	
276	7540/16	5650/15	83.6	146.6	

Out of all the tests conducted with cadavers, this was the only case of a fracture, at 50 km/h impact velocity; it was a fracture of the right hand side patella and upper part of the ilium, for a corrected load of 9400 N. No fracture of the femur was observed.

DISCUSSION

The results obtained during this study confirm that the knee-femur -pelvis combination can tolerate high loads without fractures occurring, and these loads are sometimes higher than the level allowed on a dummy by the FMVSS 208 which specifies 10,000 N.

The pulse transmitted to the buffer was determined for each knee impact. Figure 5 shows the loads recorded (maximum load) as a function of the corresponding pulse transmitted. Our load and pulse values combined are generally lower than published data (8) and, with the exception of two values, the loads remain below 10,000 N. This is due to test conditions. It would be advisable to modify them, particularly subject restraint, so that the loads obtained be greater.

A review of all the data shows that the peak load value alone is not sufficient to indicate the risk of fracture. A high knee loading must be combined with a high pulse to cause fractures. On this figure derived from (8) and completed with our data, the threshold at which fractures appear seems to be located around 200 N.s and the associated loading, in the Melvin experiments, would be in the region of 13350 N. Three subjects, with knee loadings lower than that value, sustained fractures. Two of these were accounted for by the osteoporotic condition of the subjects, as shown in the figure 5.

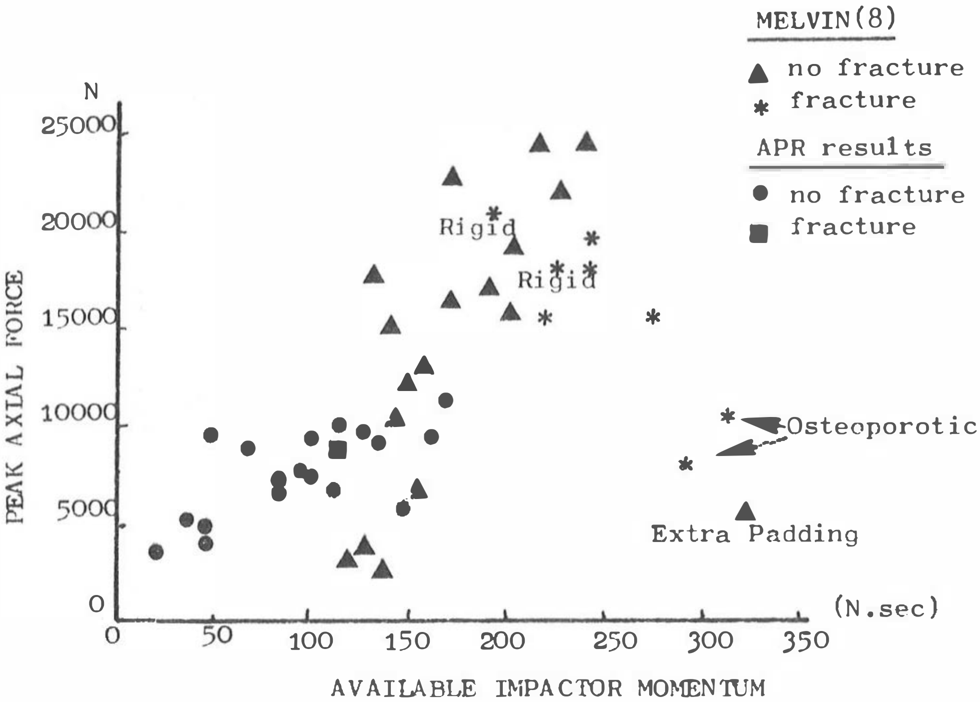


Fig. 5 -A plot of peak impact force versus available impactor momentum for axial tests.

Regarding the third case of fracture, which corresponds to one of our tested subjects, taking into account its bone condition proved necessary to try and explain this result.

Bone characterization - Preliminary tests on the bone characterisation of the subjects were carried out on the femurs sampled after impact, to try and explain the occurrence or not of a fracture at comparable pulse and load levels. The tests were initially limited to calcinations of fragments of the femurs to determine mineralization rates. The methodology was as follows:

- measurement of the total length of the femur between two flat surfaces perpendicular to the main axis of the diaphysis, one tangent to the top face of the bone head, the other to the bottom face of the condyles.

- Sampling of three fragments in geometrically different areas:

. a 3 cm-long-fragment from the middle of the overall length of the femur,
. an upper fragment of 3 cm, taken from the extreme lower limit of the trochanter minor,
. a lower fragment, 3 to 4 cm long the upper end of which is located at the start of the lower widening zone of the diaphysis and the lower end is supracondylar,

- Calcination of these fragments and determination of a mass of ashes per unit of length (C/L).

As we did not have the analysis results of all the femurs (and particularly for test No. 233 where there was a fracture), we looked for a correlation between the bone condition of the thorax and the mineralization of the femurs as determined above. Several adjustments were made and there was found to be a good relationship between the two parameters ($r = 0.93$, $n = 8$) by means of a hyperbolic adjustment. As we have the rib mineralization results of subject No. 233 (low mineralization, characteristic of osteoporotic subjects), we can obtain an estimation of the corresponding C/L value for the mineralization of this subject's femurs. The likely value thereby calculated would be 3.18 g/cm, which puts the subject the lowest in the femur mineralization scale; for the 7 other tests considered, the mean value is 4.40 g/cm with a minimum of 3.38 and a maximum of 5.49 g/cm. This result may explain why this subject sustained a fracture for a corrected load level lower than that of some other subjects.

Figure 6 compares, for the series of tests run at 50 km/h using the same deceleration pattern, the dummy and cadaver primary loads - as a function of their duration - measured at the knee stop pieces, as well as the dummy loads recorded on the femur transducers. The figure shows that knee loads are very different between the dummy and cadaver and that dummy knee and femur loads are not equivalent either.

Viano (17) wrote that the 10 kN defined by the FMVSS 208 as a top limit for dummy femur compression could be equated to contact forces on a human knee of only 5.3 to 8.3 kN. We find this result in our tests; values obtained for femur loads were 15.3 and 13.3 kN and knee loadings measured on cadavers subjected to the same tests range from 5 to 11.3 kN. We can compare these figures to the 13 kN mentioned above (8), which correspond to the occurrence of the first femur fracture for a group of unembalmed, human subjects, with short impact durations but very high contact velocities (10 to 20 m/s), and so not really meaningful or realistic for the study of impact response and tolerance characteristics of the lower members. Indeed, our experiments conducted in an automobile environment led to a knee impact velocity of about 6 m/s femur loading and impact durations for the dummy tests are approximately the same as those obtained in tests with a dummy and a production vehicle, without stop pieces and the knees hitting the vehicle instrument panel. Bearing these remarks

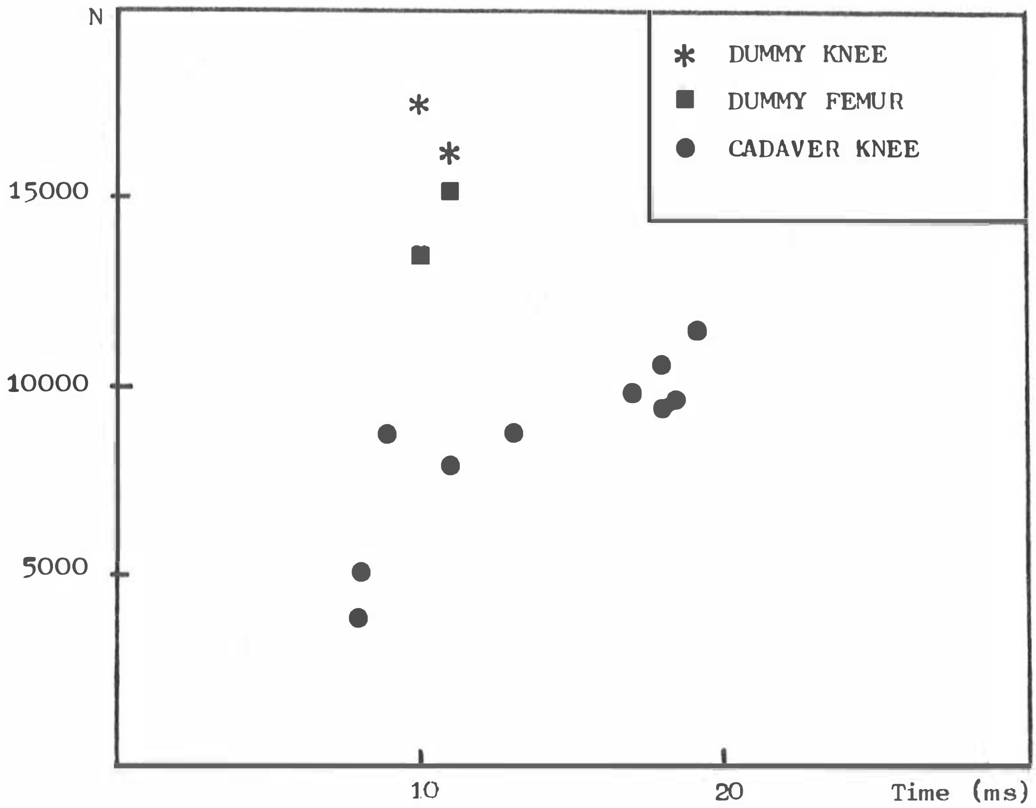


FIG. 6 . FEMUR AND KNEE LOADS FOR THE DUMMY AND THE CADAVER TEST UNDER IDENTICAL TEST CONDITIONS.

in mind, the 10 kN limit not to be exceeded on a dummy femur does not seem at all excessive.

The difference between the buffer loads recorded for the dummy and the human subjects is due to the built of the dummy which has a much greater metal skeleton mass and a structure which hardly allows any deformation of the knee area. Consequently, we can observe a load signal of a greater amplitude for the dummy though often of a shorter duration.

Furthermore, figure 7 indicates that dummy femur and knee loads are not equivalent, this being partly due to the fact that the transducer is mounted on the femur whereas knee loads are measured at the articulation of the knee. We found a mean ratio of 0.85 between the femur loads and corresponding knee loads. We can see from these few remarks that the dummy responses cannot be directly transposed to cadavers. The usual load levels (FMVSS 208) of 10 kN is not comparable to a simple knee load on a cadaver tested under the same conditions. It is therefore difficult to come to a general conclusion as to the comparative response between the dummy and the cadaver and the relationship between the values measured is too dependent upon impact condition.

CONCLUSION

An analysis of our data and comparison with other results from tests conducted in different configurations indicate that the force required to fracture the femur is very much dependent upon test conditions, and in particular on the duration of load application. Moreover, in experiments carried out with cadavers, the large variety of subject size, age and skeletal characteristics are factors which contribute to the scatter of experimental data (fractures occurring or not). Therefore, the characterization of the osseous condition of the subject is of primary importance. Now, tests are to be conducted but under more severe test conditions so as to obtain higher loads and be able to better determine the thresholds for the occurrence of fractures.

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