AN EXPERIMENTAL MODEL FOR THE STUDY OF LOWER LEG AND KNEE INJURIES IN CAR-PEDESTRIAN IMPACTS

by

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Abstract.

A theoretical analysis of the reaction forces on the human leg during car-pedestrian lateral impacts is made. The possibility to minimize these forces in the knee by varying the impact level is illustrated. A mechanical model of the lower extremity has been used in experiments simulating the leg-bumper impacts. Based on these data the significance of different bumper levels and types for lower leg and knee injuries seen in car-pedestrian accidents is discussed.

Introduction.

Earlier investigations of real and simulated car-pedestrian accidents have indicated the possibility to mitigate the injuries by modifying the car exterior (Eppinger and Pritz 1979). These conclusions were based on experiments with dummies and cadavers and were correlated to real accident data. Mathematical models of the human body during impact have been proposed (Bacon and Wilson 1976, Brun et al 1979, Padgaonkar et al 1977) and simple mechanical devices simulating certain body parts during impact have been recommended for the evaluation of the aggressiveness of various vehicle designs (Eppinger and Pritz 1979, Kramer 1979, Echavidre and Gratadour 1979).

Special interest has been focused on injuries to the lower leg and knee caused by bumper impacts and different opinions exist about the influence of the height and compliance of the bumper on the severity of injuries. According to some research groups a bumper height below 40 cm should be avoided because it may increase the impact velocity of other parts of the human body. However, other groups have recommended a low bumper level and a short bumper lead distance in combination with a smooth and compliant front profile to mitigate the leg injuries as well as the whole body injury severity (Ashton and Mackay 1979).

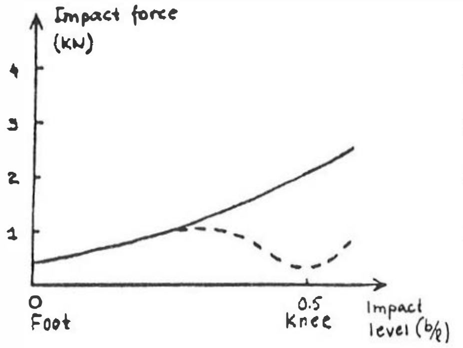
Scope.

The aims of this study was to investigate the injury mechanism and the reaction forces in the lower leg and in the knee joint during impacts to the side of the human leg.

Methods.

I. Theoretical analysis.

A theoretical analysis was made of the bumper-leg impact sequence. In this analysis a compound pendulum model was used (Appendix). The mass and dimensions of this pendulum correspond to an adult human leg. The model is based on the results from earlier experiments on human leg specimens (Aldman et al 1979). The mean bumper force at the impact point and the ligament force in the knee joint calculated on the basis of this model are illustrated in fig 1 a and b.



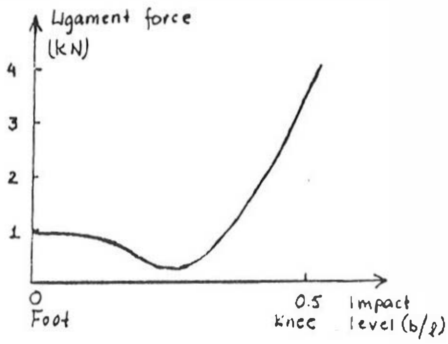


Fig 1 a. The mean bumper force for various impact levels

Fig 1 b. The mean ligament force for various impact levels

The dotted curve in fig 1 a illustrates a hypothetical deviation from the rigid model caused by a non-rigid connection of the knee joint between the more rigid thigh and lower leg. Near this joint a lower impact force should be expected when compared to the force in the rigid model. For purely lateral impacts this deviations depends on the relative bending stiffness of the leg in this direction. This bending stiffness in the non-rigid model is governed by the flexion-extension angle and a possible axial rotation of the knee, since when the knee is hyperextended or the lower leg is rotated in relation to the femur the collateral ligaments are tightened. The knee reaction forces can be considered as a combination of a shear force component and a bending force couple consisting of a ligament tension force and a joint surface compression force. An axial rotation force and an axial tensile force of the ligaments can also appear as a result of the impact. The total ligament force approximately equals the bending force plus the axial tensile and rotational forces. The axial tensile force will lower the surface compression force.

Results I.

This theoretical analysis indicates the possibility of minimizing the reaction forces on certain parts of the lower extremity during lateral impacts by chosing a suitable impact level. The bending force at the knee joint level of this model has a well defined minimum near the center of gravity of the lower leg. On the other hand the direct impact force would be higher at this level than at the knee joint level. The sum of the direct impact force and the knee reaction forces probably has a minimum value somewhere between the mid-tibia and the knee joint level. This minimum will be just above the center of gravity of the lower leg. This is illustrated in fig 2.

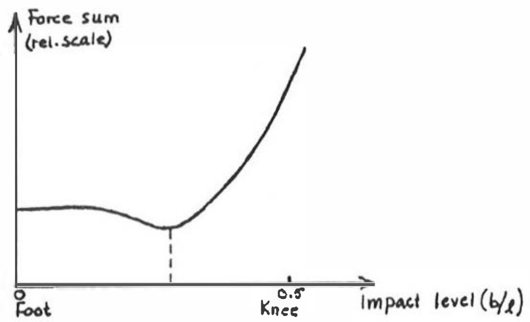


Fig 2. Hypothetical sum of impact force and knee reaction forces for varying impact levels (jointed leg).

II. The mechanical model.

A mechanical model of the adult human leg and knee was constructed. The upper and lower components are made of a turned bacelite core and the centre of gravity was adjusted to its correct position by adding sheet lead on the outside. This is then covered by a few centimetres of plastic foam to simulate the soft tissues (fig 3 a and b).

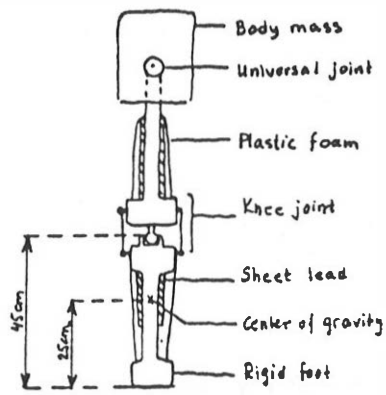


Fig 3 a. The whole leg

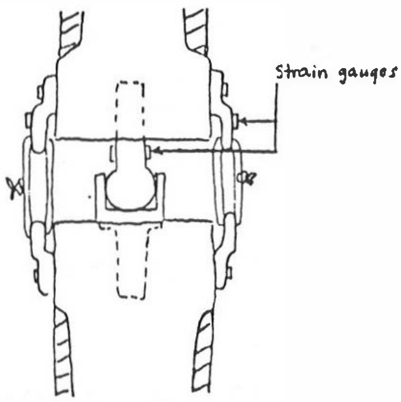


Fig 3 b. The knee joint.

The weight of the upper component is 7 kg and the lower component is 5 kg. The knee joint is simulated by a stainless steel ball and socket joint. On the attachment shaft of the ball strain gauges for shear force recordings are glued. The collateral ligaments are simulated by copper wires which are connected to the upper and lower leg components by attachment fittings with strain gauges for tensile force measurements. The collateral ligaments restrict bending of the knee in one plane. The leg is loaded by a simulated body mass. This mass is connected with the leg by an universal joint which permits a 25^{0} pendulum angle before contact is made with the side of the body mass. This angle approximately corresponds to the adduction range of the human hip joint. The leg is vertically mounted on a platform with a high friction surface and struck by a test cart carrying a rigid metal car bumper 3 cm wide (Fig 4).

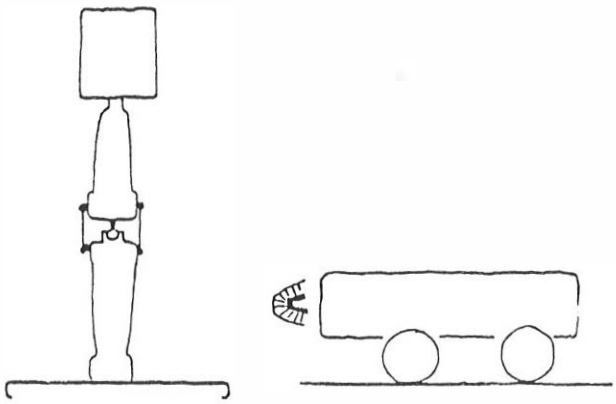


Fig 4. The experimental set up.

In some cases the bumper was covered by a 5 cm or a 10 cm thick padding of polyurethan foam. The coefficient of restitution(e) for this padding was .4 at the impact velocities used in these experiments. In two experiments a somewhat harder plastic foam material were used. The impact direction was chosen to get the maximum of ligament strain. The equipment was also instrumented with horizontal and vertical force transducers in the bumper attachment brackets and in the frame of the support platform. By this technique the tensile ligament force and the shear force on the knee joint could be recorded and compared to the bumper and ground forces at any moment during the impact.

The impact velocities, the bumper level and the paddings were varied. The impact sequence was documented by high-speed cinematography. The maximum knee deflexion angle and the angular velocity of the whole leg after impact were derived from analysis of the high-speed films.

Results II.

39 tests were made. The test data are shown in table I . At 42 cm impact level the "ligament" opposite the impact side broke in three cases (no 30, 31, 36); the impact velocities in these cases were 12, 12 and 9 km/h respectively. In another case (no 38) at 18 km/h and with a 34 cm impact level the "ligament" attachment opposite the impact side broke.

The following bumper types were used.

I.	Metal bumper;	no padding;	3 cm impact width
II.	-"-	; 5 cm polyurethan padding;	5 cm impact width
III.	-"-	; 10 cm -"-	; 5 cm -"-
IV.	-"-	; 10 cm polyvinylchloride padding;	5 cm impact width
V.	-"-	; 10 cm -"-	; 5 cm -"-

Table I. Experimental test data.

Exp No	Velocity (km/h)	Bumper level (cm)	Bumper type	Bumper force Horizontal peak (kN)	Bumper force Horizontal mom (Ns)	Bumper force Vert peak (kN) 6)	Knee Shear force peak (kN) 7)	Ligament force peak bending (kN) 8)	Ligament force peak tension (kN)	Ground force peak hor. (kN)	Ground force peak vert. (kN)	Force pulse length bumper (mS)	Force pulse length ligament (mS)	Knee distort. angle (o) 9)	Leg angular velocity γ̇ (o/sek)
22	12	12	I	4.6	12	-	-	- .9	.2	.2	.9	5	52	- 6	250
23	13	12	I	4.6	11	- .4	- .2	- .8	.1	.2	.9	5	46	- 6	270
19	12	12	II	1.1	17	-	-	-1.1	.3	.1	.8	32	50	-10	250
21	13	12	II	1.0	16	- .1	-	-1.1	.6	.1	.8	30	54	-12	250
18	12	12	III	.8	22	.0	-	-1.0	.3	.1	.4	50	70	-11	190
20	12	12	III	.7	18	0	-	-1.0	.3	.1	.7	36	52	-14	230
24	12	12	III	.8	22	- .2	+ .2	- .9	0	.1	.8	38	60	-11	240
15	12	18	I	5.6	17	- .5	-	- .7	.3	.2	.4	6	44	- 9	250
16	12	18	II	1.3	22	- .3	-	- .9	1.0	.1	.3	40	45	- 9	300
39	12	18	II	1.7	23	- .3	+ .4	- .7	0	.2	.6	28	50	- 8	370
17	12	18	III	.8	26	- .1	-	- .4	.9	.1	.4	52	90	- 6	290
14	12	24	I	5.4	24	- .4	-	+ .3	.2	.2	.5	9	10	-	-
28	13	24	I	6.0	21	-1.1	+1.3	+ .1	.3	.2	.4	7	14	+ 2	260
13	12	24	II	2.1	29	- .1	-	+ .2	1.0	.1	.3	30	30	+ 2	360
25[1)]	12	24	III	.7	20	- .2	+ .3	- .7	.1	0	0	46	62	- 4	290
26	12	24	III	1.1	26	- .2	+ .2	+ .2	.6	.1	.4	50	64	+ 4	270
27[2)]	12	24	III	1.4	35	- .1	+ .2	+ .6	.6	.2	.6	50	80	+ 7	330
12	18	24	III	1.2	35	- .2	-	+ .7	.7	0	.3	60	60	+ 6	330
1	11	34	I	3.2	13	+ .2	+1.8	-1.4	.6	-	-	7	32	+ 9	420
2	5	34	I	.8	5	+ .2	+ .8	+ .9	0	-	-	12	52	+14	-
3	9	34	I	1.6	9	+ .2	+1.3	+1.1	0	-	-	11	46	+10	3'0
4	13	34	I	2.7	12	+ .3	+1.7	+1.2	0	-	-	9	42	+16	450
5	16	34	I	4.2	17	+ .4	+1.9	+1.2	.3	-	-	8	40	+19	660
6	12	34	II	1.1	12	+ .3	+ .7	+1.0	.3	.1	.7	29	64	+14	420
7	18	34	II	2.6	19	+ .5	-+1.9	+1.2	.5	.2	.6	23	58	+17	630
9	12	34	III	.5	8	+ .2	+ .2	+1.4	.5	.2	.4	32	86	+12	380
8	18	34	III	-	-	-	-	-	-	-	-	-	-	+ 9	420
10	18	34	III	-	-	-	+ .2	+2.2	.7	-	-	-	66	+22	500
11	18	34	III	2.3	50	- .2	+ .4	+1.8	.9	.1	.7	52	76	+19	390
29	17	34	III	1.9	48	- .2	+ .6	+1.8	0	.2	.9	50	50	+28	520
32[4)]	18	34	III	1.9	43	+ .4	+ .6	+1.0	1.4	0	0	50	60	+29	520
34	18	34	IV	4.1	40	+ .3	+1.1	+ .9	.2	.1	.5	16	34	+19	540
38[3)]	18	34	V	1.9	33	- .5	+ .7	+ .7	0	0	0	30	20	+33	590
35	5	42	III	.4	19	+ .2	+ .2	+ .6	0	0	.4	80	140	-	-
36[3)]	9	42	III	.5	20	+ .2	+ .2	+ .8	0	.1	.4	60	70	+28	340
37	9	42	III	.5	22	+ .2	+ .2	+1.0	0	0	.5	80	100	+19	280
30[3)]	12	42	III	.8	23	+ .3	+ .4	+ .9	.9	0	.8	46	30	+55	280
31[3)]	12	42	III	.8	24	+ .4	+ .2	+ .8	0	0	.8	40	50	+30	360
33[5)]	12	42	III	.8	20	+ .3	+ .3	0	0	0	.9	50	0	+35	500

Remarks:
1) The leg was hanging and was not in contact with the ground.
2) An additional weight of 20 kg was added to the simulated body mass.
3) One ligament ruptured during the test.
4) A piece of wood was placed on the ground platform in such a way that only half of the "foot" was in contact with the ground.
5) The leg was rotated 90 degrees from its normal position prior to impact.
6) + indicates a lifting effect on the bumper and - that it is pressed down.
7) + indicates a shearing force on the femoral component in the impact direction and - a force in the opposite direction.
8) + indicates stretching of ligament opposite the impact side and - the ligament on the impact side.
9) + indicates knee bending in impact direction and - in the opposite direction.

The mean bumper forces for different impact levels and paddings at 12 km/h are illustrated in fig 5.

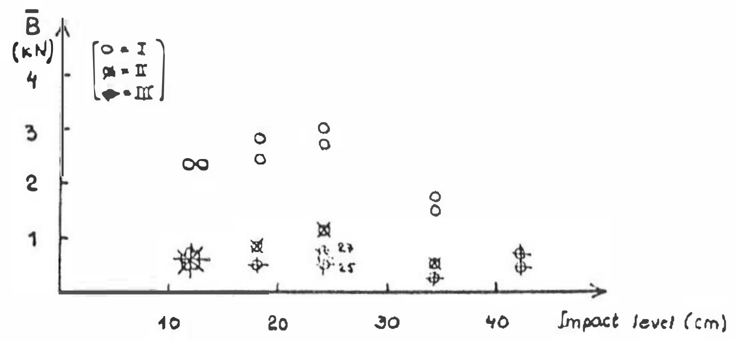


Fig 5. Mean bumper force B for different impact levels and paddings at 12 km/h.

The peak "ligament" forces for pure bending and for varying impact levels and paddings at 12 km/h are illustrated in fig 6. In this figure a ligament force value at 42 cm impact level is indicated in brackets which was calculated by linear extrapolation from experiments nr 35 and 37.

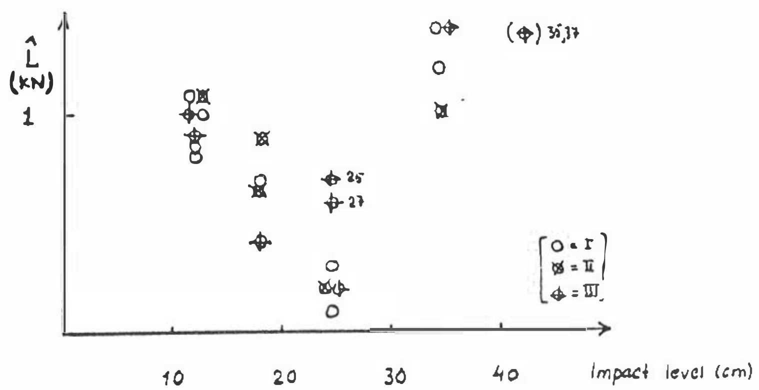


Fig 6. Peak knee ligament force L for varying impact levels and paddings at 12 km/h.

The quotient between the knee shear force H and the bumper force B for varying impact levels and paddings at 12 km/h are illustrated in fig 7.

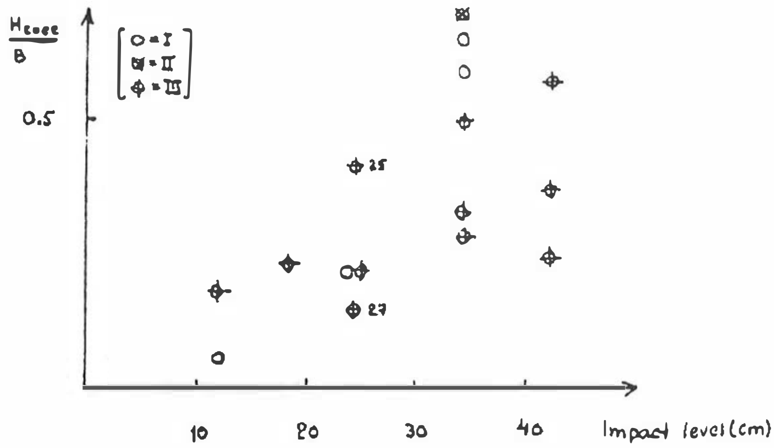


Fig 7. The quotient between the knee shear force and the bumper force for varying impact levels and paddings at 12 km/h.

The leg angular velocities after impact for various impact levels and paddings at 12 km/h are shown in fig 8.

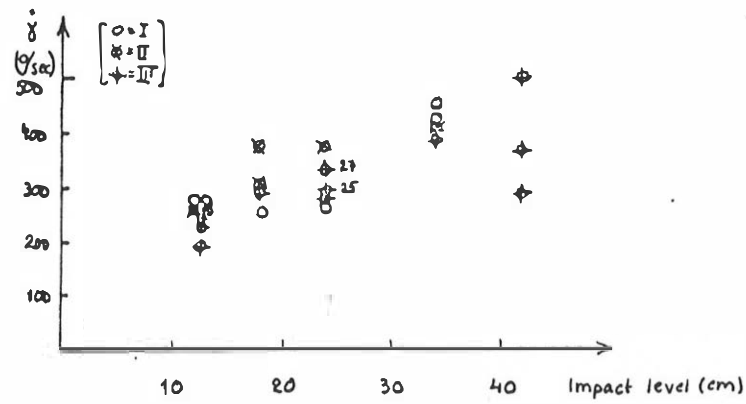


Fig 8. Leg angular velocities after impact for varying impact levels and paddings at 12 km/h.

In fig 5, 6, 7 and 8 tests no 25 and 27 are indicated because in test 25 the leg was hanging without ground contact and in test 27 an extra load of 20 kg was added to the body mass.

Discussion.

Hypothetical injury mechanisms and clinical relevance.

Impacts on the lower extremity may cause injuries to the ankle joint, the lower leg or the knee. The ankle joint injuries caused by bumper impacts often are undisplaced malleolar fractures or ligament injuries caused by indirect reaction forces due to the ground contact. The severity of these injuries were moderate in a previous study (Aldman et al 1979). These injuries occurred more often at low impact levels (25 cm).

The lower leg injuries caused by bumper impacts are soft tissue injuries and fractures at the impact level. A rigid and narrow impact surface concentrates the blow on the leg to a shorter time and a smaller area and a "high energy" fracture may occur. The skin and muscular injuries adjacent to a lower leg fracture are significant and the healing time is longer for this type of injuries compared to a "low energy" fracture (Bauer and Edwards 1965).

Knee injuries are caused by direct or indirect forces. Hypothetically the impact reaction forces will increase with increased ground friction. This influence is particularly important at low velocities at which a ligament rupture or a knee condylar compression fracture can be seen even near zero velocity. As can be seen in fig 6 (Exp no 25 and 27) the knee ligament force is only slightly influenced by varying the friction forces on the platform at moderate impact velocity. A lower bending stiffness of the leg in the impact direction at the knee level will reduce the bumper impact force and the knee shear force. If the extended leg is hit from the anterior side not far below the knee level the posterior capsule and the posterior cruciate ligament of the knee are most heavily loaded (Kennedy and Grainger 1967). If the lower leg is hit in the opposite direction the anterior cruciate ligament is maximally strained. For intermediate impact directions mixed ligamentous and menisceal injuries are possible (Kennedy et al 1974). The knee joint surface compression force also depends on the parameters mentioned above. A flexed knee seems to be more vulnerable for compression fractures compared to an extended knee according to a biomechanical study by Hirsch and Sullivan (1965). Blow fractures of the tibial or femoral condyles combined with injuries in the knee joint caused by bending, shearing and compressive forces are severe. There is a high risk for cronical joint instability and arthrosis but the long term effect of these injuries are not fully known.

Leg injuries caused by bumper impacts have been studied experimentally using human leg specimens (Aldman et al 1979). The results indicated a high knee injury risk when the leg was hit by a bumper at 45 cm level. At a 25 cm bumper level no knee injuries occurred. The AIS ratings of the injuries caused by the bumper at 45 cm impact level were significantly higher than those caused by the bumper at 25 cm level.

The results from the theoretical analysis and the experimental tests indicate that the force caused by the direct blow of the bumper will have a maximum at the center of gravity of the lower leg and a minimum at the knee level. On the other hand the knee reaction forces are minimum for an impact at the mid-tibia level and the knee reaction forces increase for an impact near knee level. This is clearly shown by the force data as well as by the cinematographic data.

An impact level where the leg injuries are least serious probably exist. It is not clear from the results presented so far what this level will be. Earlier investigations on lower leg and knee injuries caused by bumper impacts are not fully conclusive. The injury mechanisms of the various parts of the lower extremity under impact load has to be understood and the tolerance levels of the ligamentous and skeletal tissues as well as of the soft tissues has to be considered. Acceleration data for various body parts during impact are not sufficient for this purpose.

The impact force can be sufficiently reduced by incorporating deformable material in the impact area. However, a higher angular velocity of the body rotation after the primary impact may follow and result in an increased head injury risk. This has been feared if the bumper level is lowered. In the leg model an increased leg rotational velocity has not been verified for the 34 cm impact level compared to the 42 cm level. Quite the contrary at impact levels below 34 cm the angular velocity of the leg seems to diminish.

A simplified leg model has been used in this study. The shear stiffness of a real knee joint is lower. No attempt was made so far to adopt realistic elongation characteristics of the collateral ligaments. It has to be considered as a first prototype of a dummy leg for accident investigations and a more sophisticated model is scheduled. The results from this study, on the other hand indicate an optimal impact level some small distance above the center of gravity of the lower leg.

It also indicates the possibility to mitigate the direct bumper impact forces on the lower leg by incorporating a 5-10 cm energy absorbing structure to the bumper.

A synthesis of the theoretical and experimental results obtained so far with clinical facts of real car-pedestrian accidents may give some solutions for pedestrian safety problems. In order to verify these theoretical and experimental results real accident data are important. In order to arrive at an optimal car design for pedestrian protection one has to take into consideration the severity of all injuries and the long terms effect of various injuries.

Conclusions.

The bumper force has a maximum when the impact occurs near the center of gravity of the lower leg and a minimum near the knee level.

The knee reaction forces have a minimum near the center of gravity of the lower leg.

The knee ligament forces are highest for impact at the knee level.

A deformable bumper structure can reduce the bumper force but not all the knee reaction forces.

This physical model can probably be developed into an instrument for rating bumper aggressiveness in car-pedestrian accidents.

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Appendix.

A mathematical model of the human leg during lateral impact.

Approximation: The leg is represented by a homogenous rod pivoting vertically from an axis at its upper extremity. Its mass and length are known. It is struck by a force at some distance from the axis.

Problem: What are the reaction forces at the impact point and at the knee level?

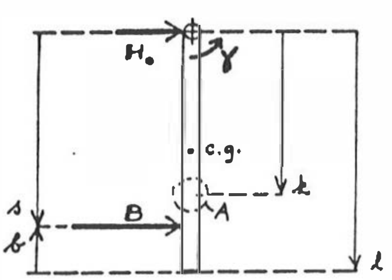

Fig A 1. Model

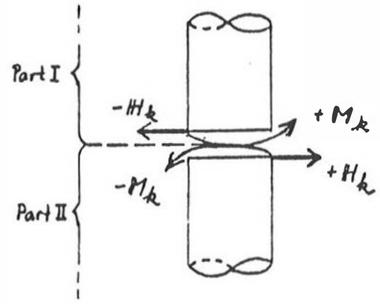

Fig A 2. Detail "A"

Definitions and presumptions.

b = bumper level (= impact level)
s = impact distance (the distance from the pivot point)
γ = rotation at the pivot point
B = impact force (= bumper force)
H_0 = reaction force at the pivot (hip joint)
H_k = transverse force at a distance k from the pivot
M_k = bending moment at a distance k from the pivot
m = the leg mass
ℓ = the leg length
I_0 = moment of inertia of the leg (rod) for the pivot point
I_k = moment of inertia of part I for the pivot point.

The impact force is calculated out of the mean angular acceleration and the deformation of the leg during impact. During the first 30 msec from impact no significant translational movement occurred of the body mass in earlier experiments on human leg specimens (Aldman et al 1979) and so the pivot point was considered at rest during this period.

Calculations.

A. The impact force and the bending moments.

The c.g. movement:

(1) $$H_0 + B = m \cdot \frac{\ell}{2} \ddot{\gamma}$$

The angular movement:

(2) $$B \cdot s = I_0 \ddot{\gamma} \text{ where } I_0 = \int_0^{\ell} \frac{m}{\ell} \cdot dr\, r^2 = \frac{1}{3} m \ell^2$$

(1), (2):

(3) $H_o = \frac{3s - 2\ell}{2\ell} \cdot B$

The c.g. movement, part I:

(4) $H_o - H_k = \frac{k}{2} \cdot \ddot{\gamma} \cdot \frac{k}{\ell} \cdot m$

(1), (2), (3), (4): If $k = \frac{\ell}{2}$ (knee)

(5) $H_{knee} = \frac{9s - 8\ell}{8\ell} \cdot B$

The angular movement, part I:

(6) $M_k - H_k \cdot k = I_k \cdot \ddot{\gamma}$ where $I_k = \frac{m}{\ell} \cdot \frac{k^3}{3}$

(5), (6): (If $k = \frac{\ell}{2}$)

(7) $M_{knee} = \frac{11s - 8\ell}{16} \cdot B$

The impact force out of (2):

(8) $B = \frac{1}{3} m \ddot{\gamma} \cdot \frac{\ell^2}{s}$

The acceleration of the impact point of the leg:

(9) $a = s \cdot \ddot{\gamma}$

Approximating $\ddot{\gamma}$ and a as constants (inelastic impact) gives us

(10) $a = \frac{v^2}{2d}$ where

v = impact velocity
d = deformation and translation of the impact point during acceleration

(8), (9), (10):

(11) $\bar{B} = \frac{1}{6} \cdot \frac{m\ell^2}{d} \cdot \frac{v^2}{s^2}$ or $B = \frac{1}{6} \cdot \frac{m}{d} \cdot \frac{v^2}{(1 - \frac{b}{\ell})^2}$ where

$\bar{B}$ = the mean impact force.

If the impact is elastic the constant in formula (11) will be $\frac{1}{3}$ instead of $\frac{1}{6}$.

With
m = 12 kg
d = .04 m
v = 3.3 m/s (12 km/h)
ℓ = 1 m
and using constant $\frac{1}{6}$ the mean impact force will be

(12) $\bar{B} \cong .5 \frac{1}{(1 - b)^2}$ (kN)

Fig C curve $\bar{B}$ illustrates the impact force for various points of impacts according to the formula (12).

B. The reaction forces in the knee.

Fig B illustrates a frontal view of the knee where the ligament and the condylar forces caused by a lateral impact on the lower leg are indicated.

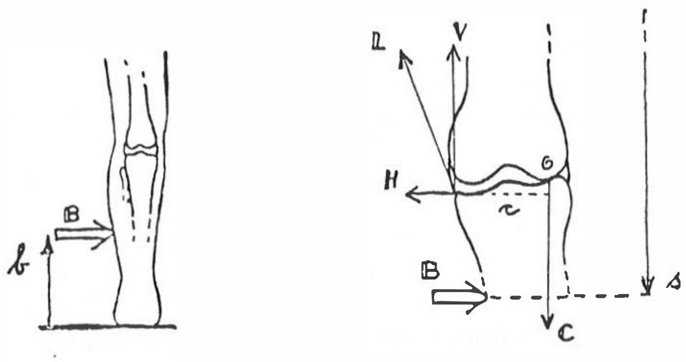

Fig B. The knee reaction forces.

In this case $s > \frac{8}{11}$ and the knee joint is forced in a varus deformation when hit from the lateral side.

The bending moment about the condylar contact point G, if the contribution from the horizontal force is neglected, is:

$M = V \cdot c$ which according to (7) gives the vertical component of the ligamentous force

$$(13) \quad V = \frac{11\,s - 8\ell}{16} \cdot \frac{B}{c}$$

The ligamentous force can be calculated by (5) and (13)

$$(14) \quad L = \sqrt{H^2 + V^2}$$

The vertical component V approximately equals the condylar force C if the knee is not pushed or pulled in axial direction.

Fig C curve L illustrates the knee ligament force $\bar{L}$ for various impact levels on the lower leg according to the formula (14) if c = .1 m.

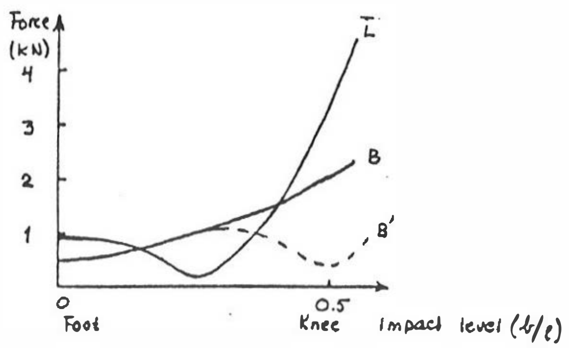

Fig C. Forces at the impact point ($\bar{B}$ = rigid, $\bar{B}'$ = jointed model) and at the knee joint ($\bar{L}$) for various impact levels.

Comments.

The bumper force $\bar{B}$ and the ligament force $\bar{L}$ of the knee are illustrated in fig C., solid curves. The dotted curve $\bar{B}'$ is hypothetically valid for a jointed leg model.

The elastic deformation of the skeleton of the leg and the lower bending stiffness of the knee have not been considered in this model. For a real leg the deformation and translation distance during impact (d) is greater in some cases. This is particularly important at the knee level. This will influence the bumper force and the knee shear force and this is to be studied in an experimental model. Hypothetically the difference between the stiff pendulum model and a jointed leg model will be minimum near the center of gravity of the lower leg where the knee bending is minimum and it will be maximum near the knee joint. This is illustrated by the dotted curve $\bar{B}'$ in fig C. Theoretically the knee shear force H_{knee} should be $.5 \cdot B < H_{knee} < 1.0 \cdot B$ when the leg is impacted at the knee level. From this follows that the direct and indirect reaction forces on the lower extremity when impacted from the side should have a minimum between the levels of the knee and the center of gravity of the lower leg.

The friction force from the ground has been ommitted in these calculations.