EXPERIMENTAL STUDIES ON LEG INJURIES IN CAR-PEDESTRIAN ACCIDENTS

by

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Abstract

A biomechanical model was used for experimental studies of leg injuries in car-pedestrian accidents. Human leg specimens including the hip joint were loaded by a concentrated body mass, balanced on an instrumented platform and impacted by a simulated car front mounted on a test cart. Injuries to the different parts of the leg specimens were analysed for various bumper levels, bumper lead angles, and bonnet edge heights at impact velocities 20 - 28 km/h. An energy absorbing and force limiting bumper was tested at 23 and 42 cm impact levels. A comparison is made of the injuries seen in real accidents and the possibility to mitigate these injuries by modifying the car front is discussed.

Introduction

The injury mechanism in real car-pedestrian accidents is sometimes obvious: lower leg and knee injuries are caused by the bumper impact, injuries in the hip and pelvis area by the bonnet edge and head injuries by the head impact on the bonnet or in the windscreen area. Other details of the impact kinematics are less obvious and this will perpetuate the discussion of how to mitigate the injuries in this type of accidents. The first impact of the car influences the kinematics of the body during the later part of the collision and thus also the injuries caused by secondary impacts. The connections between head injuries and the bumper level, the bumper lead angle and the bonnet edge height are not conclusive. Theoretically the parameters which determinate the body rotation after the primary impact should be the bumper level, the bumper lead angle and the bonnet edge height. In a mathematical simulation of the body kinematics in a car-pedestrian accident Lestrelin et al. (1980) showed a somewhat higher impact velocity of the head if the bumper level was lowered from 50 to 40-35 cm. The same was obtained if the bumper lead angle was diminished. The bonnet edge may also cause knee injuries when the body is thrown up in the air. This has been noticed in a clinical study (Bunketorp, Romanus 1981). Thus, there might be at least two possible knee injury mechanisms related to the impact of the car front.

Scope

This study was made to investigate the injury mechanisms and the tolerance level of human leg specimens impacted by a simulated car front and to correlate these results to injuries in real accidents.

Material and method

Experimental set up

An experimental model for the study of lower leg and knee injuries seen in car-pedestrian accidents has been described earlier (Aldman et al. 1979). In the previous test series human leg specimens amputated at mid-femur were used. In this study the leg specimens included also the hip joint and the lower part of the iliac bone. The specimens were loaded by a concentrated body mass, balanced on an instrumented platform and impacted by a simulated car front mounted on a test cart. In most cases the specimens were placed vertically with the knee extended and loaded by a simulated body mass. The impact angles in these cases were 70 - 80 degrees, i.e. the test cart impacted at the antero-lateral aspect of the specimens. In some cases the load was reduced to study the influence of the ground reaction forces. The following quantities were recorded and stored on magnetic tape: the impact forces on the bumper and the bonnet edge, the reaction forces on the support platform and the acceleration data from three accelerometers strapped as close as possible to the centers of gravity of the lower leg, the femur and the body mass. The impact sequence was also covered by high-speed cinematography. The experimental set up is illustrated in Figure 1.

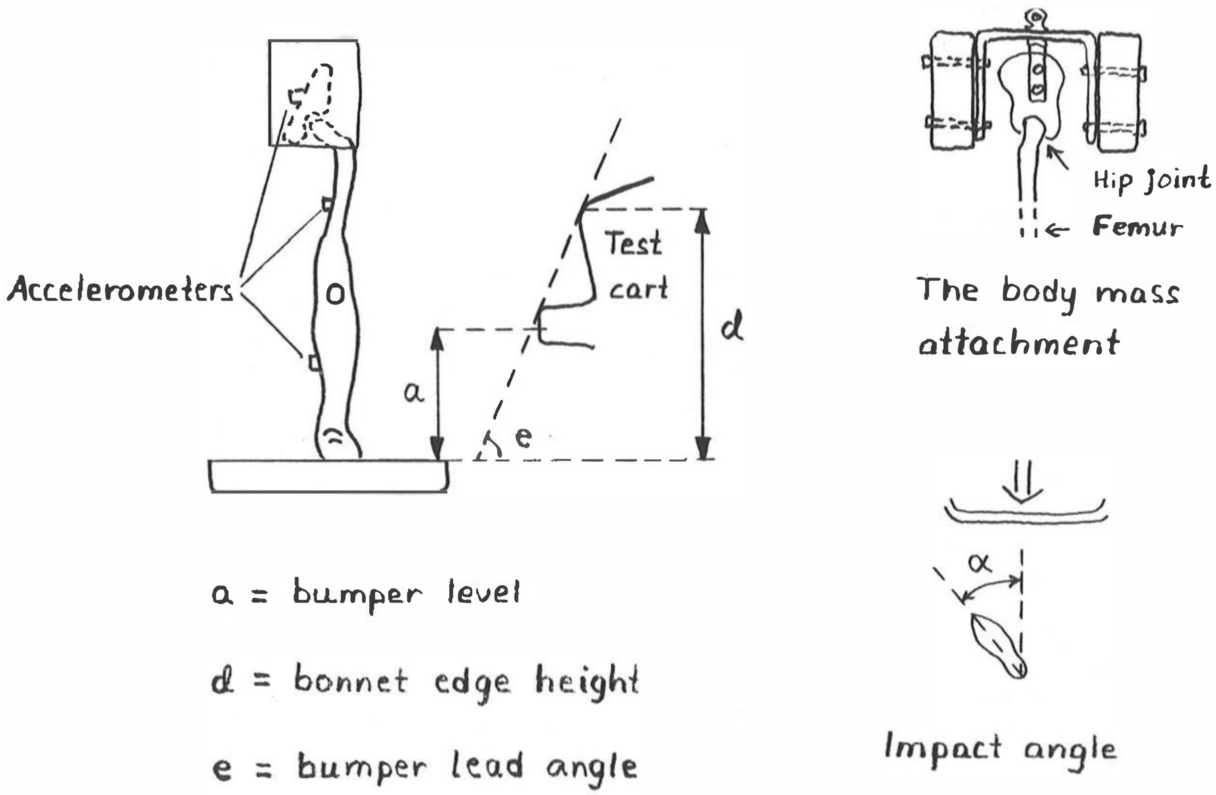

Fig. 1 - The experimental set up.

A rigid metal bumper with 3 cm impact width was used in some of the experiments. In the other a force-limiting bumper model was investigated (Fig. 2).

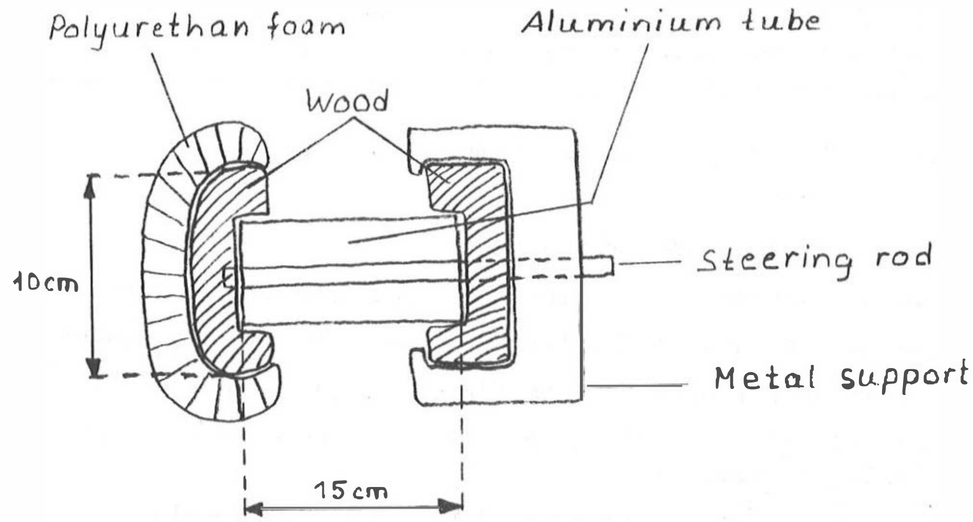

Fig. 2 - The force limiting bumper model.

The damping elements of this bumper consisted of an aluminium tube with approximately 1.5 kN maximal deformation force mounted between two wooden plates and a 2.5 cm thick polyurethan plastic foam layer covering the front plate. During impact the aluminium tube deformed plastically. The maximal deformation distance of the tube was 10 cm. The coefficient of restitution of the polyurethan foam layer was 0.4. The mass of the original metal bumper and the force limiting bumper model was 3 kg and 1 kg respectively. In some cases the bonnet was demounted to separately study the influence of the bumper impact.

Bone strength

The bone strength of the specimens were estimated from the bone mineral content (BMC), measured by dichromatic absorptiometry. In this method two monoenergetic radionuclides, ^{241}Am and ^{137}Cs, are detected simultaneously by a scintillation detector (Roos, Sköldborn 1974, Roos 1975) with intermittent scanning transversally across the specimen. The BMC is obtained by integration of the bone profile curve and is expressed in the units of grams hydroxyapatite/cm.

Scans were made at the midpoint of the lower leg (T + Fi), the tibial condyle (TC), and the midpoint of the femur (Fe) (Fig. 3). The sum of the BMC values at these three measuring points was considered approximately related to the bone strength.

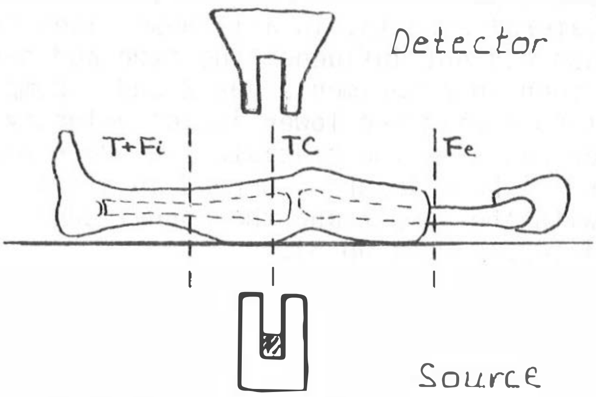

Fig. 3 - Bone mineral content measurements.

Injury classification

The instability increment of the knee joint caused by the impact was estimated. A knee instability index (KI) was calculated using the formula

$$KI = \frac{1}{n} \times \Sigma i_n \times \frac{1}{m} \times \Sigma s_m \text{ (degrees x mm; } \Sigma i_n, \Sigma s_m > 0;$$

where

i_1 = knee hyperextension difference (degrees)

i_2 = varus-valgus instability difference, knee extended (degrees)

i_3 = varus-valgus instability difference, knee flexed 30^0, (degrees)

i_4 = axial rotation difference, knee flexed 30^0, (degrees)

s_1 = sagittal instability difference, lower leg outward rotated, knee flexed 30^0 (mm)

s_2 = sagittal instability difference, lower leg inward rotated, knee flexed 30^0 (mm)

s_3 = sagittal instability difference, lower leg neutral, knee flexed 30^0 (mm).

The specimens were radiographed before and after the tests and dissected by one of the orthopaedic surgeons who was the principal investigator of the real car-pedestrian accidents (O.B.).

If no significant injuries were noted in the knee or the lower leg the specimens were used a second time.

Statistical tests were made with the χ^2-method using Yates correction for small samples (Maxwell 1961).

Results

Nineteen leg specimens were used in 24 experiments. They originated from individuals 47 - 92 years of age. Tables 1, 2, and 3 show the test conditions, the kinetic data and the injuries.

Rigid bumper at 42 cm level with prominent bonnet edge (Nos 1 - 5)

The rigid bumper was used in experiments nos 1 - 5 at 42 cm level. This corresponds to 6 - 9 cm below the knee joint. The bumper lead angle was 85 degrees, the bonnet edge height 71 cm, i.e. approximately at the mid-femur level and the impact velocity 20 - 28 km/h. In experiments nos 1 - 3 a low body mass was used (23 kg). In the other two tests (nos 4 and 5) the body mass was 71 kg. The impact angle was 70 - 80 degrees, i.e. the specimens were impacted at the anterolateral aspects. In all these cases severe knee injuries were noted. The body mass did not influence the type and severity of the injuries as can be seen in experiments nos 2 and 3 compared to 4 and 5. The knee injury in test no 1 with the lower impact velocity 20 km/h was less severe than in the other cases as the cruciate ligaments were not ruptured but the maximum AIS for the knee injuries were 3 in all these experiments. The injury mechanisms with the bumper and the bonnet edge in this set up are illustrated in Figure 4 (experiment no 5).

TABLE 1 TEST CONDITIONS

Exp No	Spec No	Specimen: Age, sex	Specimen: Weight kg	Specimen: Tibia lenght cm	Specimen: Tibia condyle width mm	Specimen: Bone mineral content (BMC) g/cm	Specimen: Simul. body weight kg	Specimen: Impact angle (α) deg.	Bonnet edge height cm	Bumper Lead angle [3] deg	Bumper Level cm	Bumper Type [4]	Velocity km/h
1	40	68m	5.6	38	88	11.7	23	70	71	85	42	R	20
2	41	87m	6.0	37	88	16.7	23	80	71	85	42	R	25
3	42	75m	5.5	37	81	15.1	23	70	71	85	42	R	25
4	43	79f	5.5	37	80	9.4	71	70	71	85	42	R	27
5	44	61m	6.5	39	96	15.1	71	70	71	85	42	R	28
6	49	81f	4.5	33	76	10.8	26	70	71	60/75	42	D	26
7	53	79m	6.7	38	88	15.7	26	70	71	60/75	42	D	25
8	54	77m	6.3	39	84	15.4	45	70	71	60/75	42	D	25
9	57	78m	4.5	35	79	12.5	45	70	71	75/90	23	D	26
10	58	90m	6.5	42	88	17.0	45	70	71	75/90	23	D	26
11	55	75m	5.6	37	84	13.0	45	70	52	60/75	23	D	26
12	56	89f	5.0	35	76	10.9	45	70	52	60/75	23	D	26
13	59	92f	5.3	36	86	18.3	45	70	82	45/50	23	D	26
14	60	47m	5.7	34	82	11.8	45	70	82	45/50	23	D	26
15	45	86m	5.5	37	87	9.6	71 [1]	70	-	-	23	D [5]	27
16	46	77f	6.2	39	81	10.9	71	70	-	-	23	D [5]	26
17	47	69m	5.5	32	84	14.3	71	70	-	-	23	D	26
18	48	82m	5.5	39	97	12.3	71	70	-	-	23	D	26
19	50	84f [6]	4.5	37	82	13.1	26 [2]	0	71	60/75	42	D	25
20	51	86m	5.5	37	87	9.6	26 [2]	0	71	60/75	42	D	25
21	52	66m	5.5	32	84	14.3	26 [2]	0	71	60/75	42	D	25
22	61	92f	5.3	36	86	18.3	45	70	71	60/75	42	D	26
23	62	75m	5.6	37	84	13.0	45	70	71	60/75	42	D	26
24	63	47m	5.7	34	82	11.8	45	70	71	60/75	42	D	26

1) Load on platform: 45 kg
2) Load on platform: 5-10 kg, knee flexed 45 degrees
3) In experiments no 6-24 : values with the aluminium tube intact and compressed
4) R = rigid metal bumper, D = deformable bumper
5) Without padding
6) Knee contraction

Table 2 Kinetic data

Exp No	Spec No	Impact force: Bonnet edge (KN)	Impact force: Bumper (KN)	Leg rotation: Velocity at 200 ms (deg/s)	Leg rotation: Max angle (deg)	Acceleration: Lower leg (g)	Acceleration: Femur (g)	Acceleration: Body mass (g)
1	40	1.2	1.1	275	150	55	110	-
2	41	1.9	2.9	350	-	105	115	30
3	42	1.9	2.6	300	110	95	112	15
4	43	-	-	325	160	-	-	-
5	44	2.2	2.9	225	170	115	125	35
6	49	1.6	1.2	210	170	80	45	15
7	53	0.9	1.5	370	130	80	-	10
8	54	2.9	1.1	480	170	110	-	10
9	57	7.3	1.1	210	170	>100	95	-
10	58	5.0	1.4	-	120	80	75	-
11	55	3.4	1.6	400	-	100	-	20
12	56	3.1	1.4	325	140	>100	-	20
13	59	6.4	1.5	275	140	90	-	<5
14	60	4.0	1.1	275	120	90	-	5
15	45	-	0.9	175	120	165	70	<5
16	46	-	1.0	140	130	95	70	20
17	47	-	1.2	250	90	105	105	10
18	48	-	1.3	250	90	100	40	15
19	50	-	1.0	250	170	60	90	15
20	51	0.9	0.5	-	-	55	-	20
21	52	1.7	0.6	340	170	25	-	25
22	61	2.8	1.1	650	170	-	55	10
23	62	3.8	1.3	525	170	-	65	10
24	63	3.9	1.4	-	-	-	70	30

Table 3 Injuries (AIS)

Exp No.	Spec No.	Knee: Fracture: Fem cond	Knee: Fracture: Tib cond	Knee: Ligament inj.: Medial	Knee: Ligament inj.: Laral	Knee: Ligament inj.: Ant cruc	Knee: Ligament inj.: Post cruc	Lower leg: Fracture: Tib	Lower leg: Fracture: Fib	Ankle joint: Fr	Ankle joint: Lig inj	Knee instability (KI)	Knee injury cause: Bumper	Knee injury cause: Bonnet edge	Comments
1	40	0	0	3	0	0	0	0	2	0	0	-	+	+	
2	41	0	0	3	0	3	3	0	0	0	0	-	+	+	
3	42	0	0	3	0	3	2	0	0	0	0	5.0	+	+	
4	43	0	0	3	0	3	3	0	2	2	0	10	+	+	
5	44	0	3	3	0	0	0	0	0	0	0	3.9	+	+	
6	49	0	0	3	0	0	3	0	0	0	0	-	+	+?	
7	53	3	0	0	0	0	0	0	0	0	0	2.8	+	+?	
8	54	0	0	3	0	0	2	0	0	0	0	1.4	+	+?	
9	57	0	0	2	3	0	2	0	0	2	2	-	-	+	
10	58	0	0	3	0	0	0	0	0	0	0	2.3	-	+	
11	55	0	0	-	-	-	-	0	0	0	0	0.4	-	+	
12	56	0	0	3	3	0	3	0	0	0	0	2.2	-	+	
13	59	0	0	-	-	-	-	0	0	0	0	<0.1	-	-	
14	60	0	0	-	-	-	-	0	0	0	0	<0.1	-	-	
15	45	0	0	-	-	-	-	0	0	2	0	0.2	+	-	
16	46	0	0	0	0	0	0	3	3	0	0	<0.1	+	-	
17	47	0	0	-	-	-	-	0	0	0	0	<0.1	-	-	
18	48	0	0	0	0	0	0	3	3	0	0	-	+	-	
19	50	0	3	0	0	0	0	0	0	0	0	-	+	-	
20	51	0	3	0	0	0	0	0	0	0	0	3.5	+	-	=15
21	52	0	0	0	3	0	2	0	0	0	0	0.8	+	-	=17
22	61	0	0	0	0	0	3	0	0	0	0	0.2	+	-?	=13
23	62	0	3	0	0	0	0	0	0	0	0	3.0	+	-?	=11
24	63	0	0	0	0	0	2	0	0	0	2	0.2	+	+?	=14

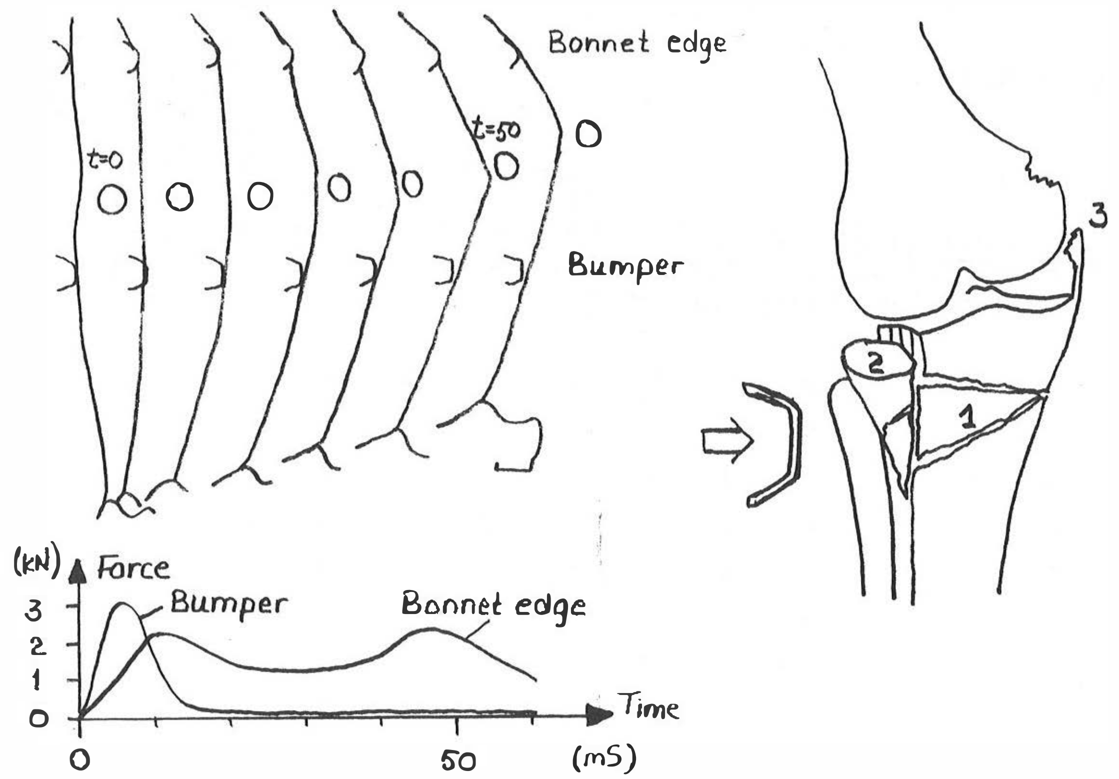

Fig. 4 - Kinetics and injuries in experiment no 5.

The probable injury sequence in this case was

1 - A transverse fracture below the tibial condyles at the impact level caused by the rigid bumper at 5 mS.

2 - A split and depression fracture of the lateral tibial condyle caused by the bumper and/or the bonnet edge at 10 mS.

3 - An avulsion of the femoral attachment of the medial collateral ligament caused by the bonnet edge at 45 mS.

The injuries 2 and 3 might have occurred in the opposite order but in that case the cruciates should have been more strained than was seen at the dissection.

Deformable bumper at 42 cm level with less prominent bonnet edge (Nos 6 - 8)

In experiments nos 6 - 8 the deformable bumper was used at 42 cm level. The bonnet edge height was the same as in the earlier experiments. The bumper lead angle was $60^{0}/75^{0}$ i.e. 60 degrees with the tube intact and approximately 75 degrees with the tube compressed. The specimens were loaded with 26 or 45 kg simulated body weight. In these cases somewhat less severe knee injuries were noted compared to the experiments nos 1 - 5. They were caused by the bumper but also by the bonnet edge.

Deformable bumper at 23 cm level with varying bumper lead angles and bonnet edge heights (Nos 9 - 14)

In experiments nos 9 - 14 the deformable bumper was used at 23 cm level. The body mass was 45 kg. The bumper lead angle and the bonnet edge height varied. In tests nos 9 and 10 these parameters were $75^{0}/90^{0}$ and 71 cm respectively. The knee injuries in these cases were severe (AIS max = 3) and caused by the bonnet edge.

In tests nos 11 and 12 a lower bonnet edge (52 cm) and a smaller bumper lead angle ($60^\circ/75^\circ$) were used. In one of these (no 12) severe knee injuries occurred. In the other (no 11) a moderate knee injury was caused as indicated by the knee instability index. The injury cause in these two cases was the bonnet edge. In tests nos 13 and 14 the bumper lead angle was $45^\circ/50^\circ$ and the bumper height 82 cm. No injuries were noted in these cases.

Deformable bumper at 23 cm level without bonnet (Nos 15 - 18)

The deformable bumper was also used in experiments nos 15 - 18. The bumper level was 23 cm and the simulated bonnet was demounted. The body mass was 71 kg but in test no 15 the load on the platform was reduced to 0.45 kN by lifting the leg. In two of these tests (nos 16 and 18) similar types of lower leg fractures were seen. In one other test an undisplaced fracture of the medial malleolus was noted. In the other there was no injury at all. The force limiting effect of this bumper and the lower leg fracture type are illustrated in Fig. 5.

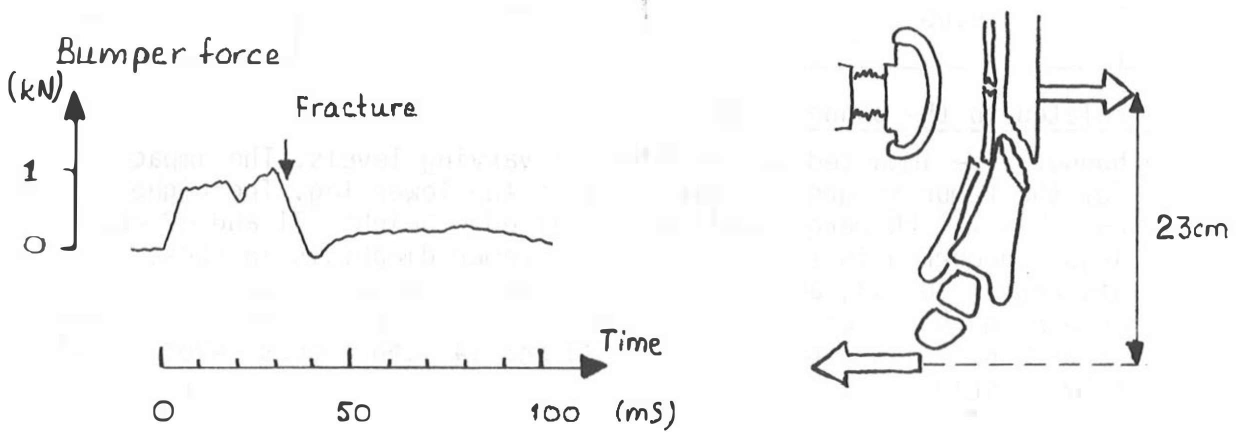

Fig. 5 - Bumper force and injury in test no 16.

Special experiments (Nos 19 - 24)

In experiment no 19 the leg specimen had a knee contraction of 45 degrees. The specimen was mounted with a low load on the platform and impacted at the anterior aspect. The test cart was as in experiments nos 6 - 8. A severe compression fracture of the tibial condyle was noted. In tests nos 20 and 21 specimens from earlier tests were used with the same experimental set up. The same type of fracture was seen in test no 20. In test no 21 posterior knee ligament injuries were caused. In tests nos 22 - 24 specimens from earlier tests were used with the same experimental set up as in tests nos 6 - 8. Knee injuries were seen in all cases. In test no 24 only a moderate knee injury was noted. This specimen originated from a 47 year old man.

Injuries AIS related to bumper level, bumper lead angle and bumper type

For anterolateral impacts ($\alpha = 70 - 80^\circ$) the influence of the bumper level, the bumper lead angle and the bumper types is illustrated in Table 4.

In experiments nos 9, 10 and 12 the AIS = 3 injuries were located to the knee joint. They were caused by the bonnet edge and not by the bumper. These cases were omitted in this analysis. The 23 cm bumper level caused less severe injuries than the 42 cm level. A deformable bumper and a bumper lead angle below $60^\circ/75^\circ$ also seemed to cause less severe injuries.

Table 4 - Injuries related to bumper level, bumper lead angle and bumper type for the different experiments

Injuries Max AIS	Bumper level (cm) 23	42	Bumper lead angle (degrees) <60/75	$\geq$60/75	Bumper type D	R
2	11,13,14, 15,17	24	13,14, 15,17	11,24	11,13, 14,15, 17,24	-
3	16,18	1,2,3,4, 5,6,7,8, 22,23	16,18	1,2,3,4, 5,6,7,8, 22,23	6,7,8, 16,18, 22,23	1,2, 3,4, 5
$P(\chi^2)$	<0.05		<0.15		<0.20	

Injuries related to the bonnet edge

The bonnet edge impacted on the femur at varying levels. The impact tolerance of the femur is greater than that of the lower leg. The highest bonnet edge forces 4.0 - 7.3 kN were noted for bonnet edge heights 71 and 82 cm. Still no injury occurred to the hip or to the femur diaphysis in these cases (experiments nos 9, 10, 13, and 14). In experiments nos 9 and 10 knee injuries were caused during the later part of the impact by the prominent bonnet edge. No injuries were noted in experiments nos 13 and 14 with bumper level 23 cm and bumper lead angle $45^0/50^0$.

Leg rotation after impact

The angular velocity of the specimen 200 mS after impact and the maximal rotation angle was analysed for varying bumper levels and bonnet heights (Tables 1 and 2). The maximal leg rotation angle caused by the deformable bumper and bonnet edge was compared for 23 and 42 cm bumper level. This was greater at the higher bumper level but the difference was not statistically significant. The influence of the bonnet edge height and the bumper lead angle was not obvious.

Injuries and accelerometer data (Table 5)

Table 5 - Accelerometer data of the lower leg for the different experiments

Injuries AIS	Lower leg acceleration <100 g	$\geq$100 g
0 or 1	3,6,7,10, 11,13,14	2,5,8,12,17
2 or 3	1,16	9,15,18
$P(\chi^2)$	p>0.5	

Higher accelerations of the lower leg are not well correlated to higher leg injury severity. The accelerations of the femur and lower leg were not correlated to the knee injury severity at all.

Comparison with injuries seen in real accidents

In the real accident study knee and lower leg injuries were seen in approximately 90%. Knee ligament injuries were as usual as intra-articular fractures of the knee. The type and severity of the knee injuries did not differ from the injuries seen in the experimental model. A static bumper level below 40 cm did not cause severe knee injuries in the real accidents. The tibia fractures were located 0 - 13 cm below the static bumper level. The walking impairment at clinical follow-up tended to be less for bumper levels below 40 cm. Walking impairment was seen even below 30 km/h impact speed in several cases. The bumper level and the bumper lead angle did not influence the presence of head injuries. Injuries to the femur and hip were caused by the bonnet edge and to the ankle joint by the bumper. The number of cases with injuries in these regions were too small to allow statistical analysis.

Discussion

The results from this experimental study indicate a high risk for knee injuries caused by the bumper and bonnet edge with ordinary front profiles. Knee injuries are noted to a great extent even below 30 km/h impact velocity and they are not correlated to the leg accelerations. Knee injuries mostly are the result of a direct bumper impact but also caused by the bonnet edge during the later part of the collision. The influence of the bonnet edge height on the injuries to the lower extremity is not clear. This parameter does not seem to influence the risk for leg injuries if the bumper lead angle is below 60 degrees. However, pelvic injuries seem to be frequently caused by the bonnet edge. The bonnet edge height probably influences the risk for injuries to the head, neck and upper torso in an impacted child and should not primarily be evaluated from leg injury protection criteria for adults.

The bumper also causes fractures of the lower leg and injuries to the ankle joint. These injuries are correlated to the impact force, the impact velocity and probably the acceleration. During the stance phase in walking and if the ground friction is high the risk for fractures at or just below the impact level will increase. The bone mineral content in those specimens which fractured at the lower leg were somewhat lower than the mean value (13.4 g/cm) of all the specimens. This fact alone cannot explain the fractures in these cases. The specimen used in experiment no 15 had a lower bone mineral content but no lower leg injury was produced. The most important injury mechanism at 23 cm bumper level is probably the combination of the bumper impact force and the ground reaction force from a heavily loaded leg. A reduced energy transmission to the lower leg is important to avoid open fractures with extensive soft tissue damage and impaired fracture healing. This is possible to achieve with an elastic bumper with a sufficiently large impact area as has been shown in an earlier experimental study (Aldman et al. 1979). On the other hand an elastic bumper will cause a rebound movement of the leg after the impact and this may rise the risk for other injuries during the later part of the impact sequence. An energy damping and force limiting bumper which deforms plastically during impact will not behave in this way and will reduce the risk for "high energy injuries". The optimal deformation characteristics of this bumper cannot yet be determined from our results. The capacity of bone to absorb energy during impact depends on the visco-elastic properties of the bone(Dumbleton, Black 1975). The critical force limit for lower leg fractures 4.3 kN as indicated by Kramer et al. (1973) and 3.7 kN as indicated in our earlier investigation (Aldman et al. 1979) are misleading as they are valid for rigid bumpers. The strain rate for maximal energy absorption without fracture should be determined for lower legs before a bumper force limit can be recommended. This limit might be lower than 1 kN as indicated in this study.

The car front design should be determined by protection criteria against serious injuries at the time of accident as well as against injuries causing permanent impairment. A static bumper level above 40 cm and a bumper lead angle greater than 60 degrees should be avoided. An optimal impact level between 25 and 30 cm for minimal knee and lower leg injuries was suggested in a theoretical and mechanical analysis of leg impacts (Aldman et al. 1980). The static bumper level was 0 - 13 cm above the fracture centres in the clinical study. Thus, the optimal static bumper level should be 27.5 + 6.5 = 34 cm for adult pedestrians. This is in good agreement with Stürtz (1981).

Conclusions

This biological model system can simulate the kinetics and disclose the injury mechanisms of the lower extremity impacted by a vehicle front in real car-pedestrian accidents.

A bumper located at or just below the knee level and a prominent bonnet edge are both correlated with increased risk for knee injuries.

A force limiting and energy-absorbing bumper impacting at the lower half of the tibia and a bumper lead angle below 60 degrees will reduce the risk for knee injuries.

High ground friction will raise the fracture risk during the stance phase of walking if the leg is impacted at lower bumper levels. The fracture tolerance limit may be below 1 kN in these cases.

Fractures of the femur diaphysis and the hip joint are not usual below 30 km/h impact velocity.

Lowering of the impact level from 42 to 23 cm will reduce the severity of the leg injuries and not increase the leg rotation after impact.

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References

Aldman, B., Lundell, B., Thorngren, L., Bunketorp, O., Romanus, B. (1979). Physical simulation of human leg-bumper impacts. Proc of the 1979 IRCOBI Conference on the Biomechanics of Trauma, Göteborg, September 1979, pp 232 - 242.

Aldman, B., Thorngren, L., Bunketorp, O., Romanus, B. (1980): An experimental model system for the study of lower leg and knee injuries in car--pedestrian accidents. Proc of the 1980 IRCOBI Conference on the Bio--mechanics of Impacts, Birmingham, September 1980, pp 180 - 193.

Bunketorp, O., Romanus, B. (1981) Clinical studies on leg injuries in car--pedestrian accidents. Proc of the 1981 IRCOBI Conference on the Bio--mechanics of Impacts, Salon de Provence, September 1981.

Dumbleton, J.H., Black, J. (1975): An introduction to orthopaedic materials. Charles C. Thomas, Springfield, Illinois.

Kramer, M., Burow, K., Heger, A. (1973) Fracture mechanism of lower legs under impact load. Proc of the 17th Stapp Car Crash Conference, Oklahoma, November 1973, pp 81 - 100.

Lestrelin, D., Brun-Cassan, F., Fayon, A., Tarriere, C., Castan, F. (1980): "Prakimod": Mathematical simulation of accident victims. Validation and application to car-pedestrian collisions. Presented at the 8th ESV--Conference, Wolfsburg, October 1980.

Maxwell, A.E. (1961) Analysing qualitative data. Methuen, London, pp 163.

Roos, B., Sköldborn, H. (1974): Dual photon absorptiometry in lumbar vertebra. I. Theory and Method. Acta Radiologica, Vol. 13, Fasc. 3, June 1974.

Roos, B. (1975): Dual photon absorptiometry in lumbar vertebra. II. Precision and Reproducibility. Acta Radiologica, Vol. 14, Fasc. 3, June 1975.

Stürtz, G. (1981) Das Kind in Verkehrsunfall: Biomechanik, Anforderungen an die äussere und innere Sicherheit von Kraftwagen. VDI-verlag, Düsseldorf.