Respons of the Knee Joint in Lateral Impact:

Effect of Shearing Loads

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ABSTRACT

During the 80's a new type of crash impact dummy, the rotationally symmetrical pedestrian dummy (RSPD), suitable for the assessment of car front aggressiveness in pedestrian impacts was developed (Aldman, 1985). This dummy enables measurement of biomechanical parameters, such as moments and forces at the knee joint level which are related to the injury mechanisms.

To determine the ultimate resistance to shear force or bending moment of the human knee, it was desirable to make separate experiments, where only one of those two parameters affects the biological material at the time.

In this study an experimental method for assessment of the shearing force in the lateral direction at the knee joint has been developed. The maximum shearing force in the lateral direction the knee could bear without injuries was determined. Injuries were described by measurements of the knee laxity and by dissection of the knee region.

Nineteen tests with human cadaver legs were carried out under dynamic conditions, nine at a velocity of 15 km/h and ten at 20 km/h. The results show the necessity of discussing two different injury mechanisms.

The first injury mechanism, which occurs at about 5 milliseconds after impact, is directly correlated with the force generated by the local acceleration of the biological system. The consequences of this force are injuries at the contact point and extra-articular injuries. The mean peak force correlated with this injury mechanism was 180 (±38) daN for an impact velocity of 15 km/h and 257 (±45) daN for an impact velocity of 20 km/h.

The second injury mechanism, which occurs at about 15 - 20 milliseconds after impact is correlated with the force transferred through the knee joint when the thigh was accelerated. The consequences of this force are intra-articular injuries of the knee joint. The mean peak force correlated with this injury mechanism was 257 (±37) daN at impact velocity of 15 km/h and 322 (±46) daN at impact velocity of 20 km/h.

1. INTRODUCTION

The major part of the accidents, in which a pedestrian is hit by the front structures of a passenger car, seems to take place in urban areas when the pedestrian enters a traffic lane in an attempt to cross the street more or less perpendicularly. Since most adult pedestrians in this type of accidents are elderly people, their own velocity is quite low relative to that of the car (Appel et al., 1975; Lestrelin et al., 1985). The first contact between the two occurs when the most protruding part of the car front, usually the bumper, hits the pedestrian's lower extremities from the side at the knee level.

Improvement of pedestrian safety is considered a priority in crash injury mitigation. Protection of pedestrians in collisions with cars is a complex problem. Analysis of this kind of accidents show that two body segments are overrepresented in the injury statistics, the lower limbs and the head. Ashton and Mackay (1979) reported that leg injuries were represented in about 60% of non-fatal injuries.

European Experimental Vehicles Committee in EEVC/CEVE (1982) extensively reported that injuries to the leg are one of the most common forms of trauma associated with pedestrian accidents.

The contact with the car bumper and the subsequent acceleration of the pedestrian leg result in a rather complex injury mechanism. The knee joint is subjected to bending moment and to shear force. The effect of this is compressive loads on the nearest tibia condyle and tensile forces in the ligaments and the joint capsule. The inertia of the foot causes the lower leg to rotate, resulting in shear in the soft tissues in contact with the bumper and possibly also torque transmitted to the knee region. It seems that the worst case would be when the bumper first impacts the leg supporting the body and that this impact occurs close to the super-extended knee joint (Kajzer, 1989).

During the 70's and the 80's many research groups designed experimental car fronts with the aim to protect pedestrians. Testing of these structures has often been made with traditional anthropometric test dummies (Hybrid II, Eurosid) or human cadavers.

However, EEVC/CEVE (1982) analysis of currently available evaluation methods shows that present anthropometric test dummies are not able to give a human-like and repeatable impact response in pedestrian tests.

When normal or only slightly modified anthropometric test dummies (Hybrid II, Eurosid) have been used for simulation of car to pedestrian impacts it has so far been customary to place them with their feet separated and the dummy weight more or less loading both legs. This seems to represent a very rare accident situation. It does not represent the worst case as far as the risk for leg injuries is concerned and this position of the legs may in some cases induce a rotation about the longitudinal axis of the body.

When cadavers have been used in simulations of car to pedestrian impacts the body was usually suspended and released only shortly before each test. In such tests the body attitude varied and the legs were usually not loaded with the knees in a super-extended position as they are most of the time in walking.

Several researchers performed studies of femur tolerance to axial static or dynamic compression (Messerer, 1880; Powell et al. 1974 and 1975; Viano and Stalnaker, 1980). Patella fracture tolerance was reported by Patric et al. (1966, 1967), Melvin and Stalnaker (1976) and Stalnaker et al. (1977). Studies of the static bending strength of the tibia and the femur were summarized by Yamada (1970) and Kramer et al. (1973) presented data from 209 tests under dynamic conditions. Knee joint tolerance in 90° femur-to-tibia orientation was investigated by Viano et al. (1978). All this information is not directly useful in mechanical or mathematical models for simulation of the pedestrian leg in lateral impact.

Tests reported by Kajzer (1989) showed that lowering the level of the first impact to a point near the centre of gravity of the lower leg reduced the bending moment over the knee joint to almost zero. The shear force on the other hand was influenced by the compliance of the structure in first contact with the leg. The development of the RSPD made it possible to assess bending moment and shear force separately.

For this reason the Department of Injury Prevention at Chalmers University of Technology in Göteborg and the Laboratory of Chock and Biomechanics at INRETS in Bron decided to make special studies in order to determine the influence of shearing and bending effects on the knee joint.

These experimental studies were made at the Laboratory of Applied Biomechanics (LBA) of the Faculte of Medicine in Marseille, France and was financially supported by the Swedish Transport Research Board (TFB) and the Institut National de Recherche sur les Transports et leur Securite (INRETS) in France.

The first phase of the project started with a description of the effects from shear load of the human knee region in a lateral direction.

2. METHODOLOGY

2.1 TECHNICAL PART

Dynamic testing of the leg was made using an impactor which was propulsed by sandows. The mass of the mobile part of the impactor was approximately 40 kg. A specially designed impact arm equipped with force transducers was mounted in front of the impactor. Initial test conditions for our tests are shown in Figure 1.

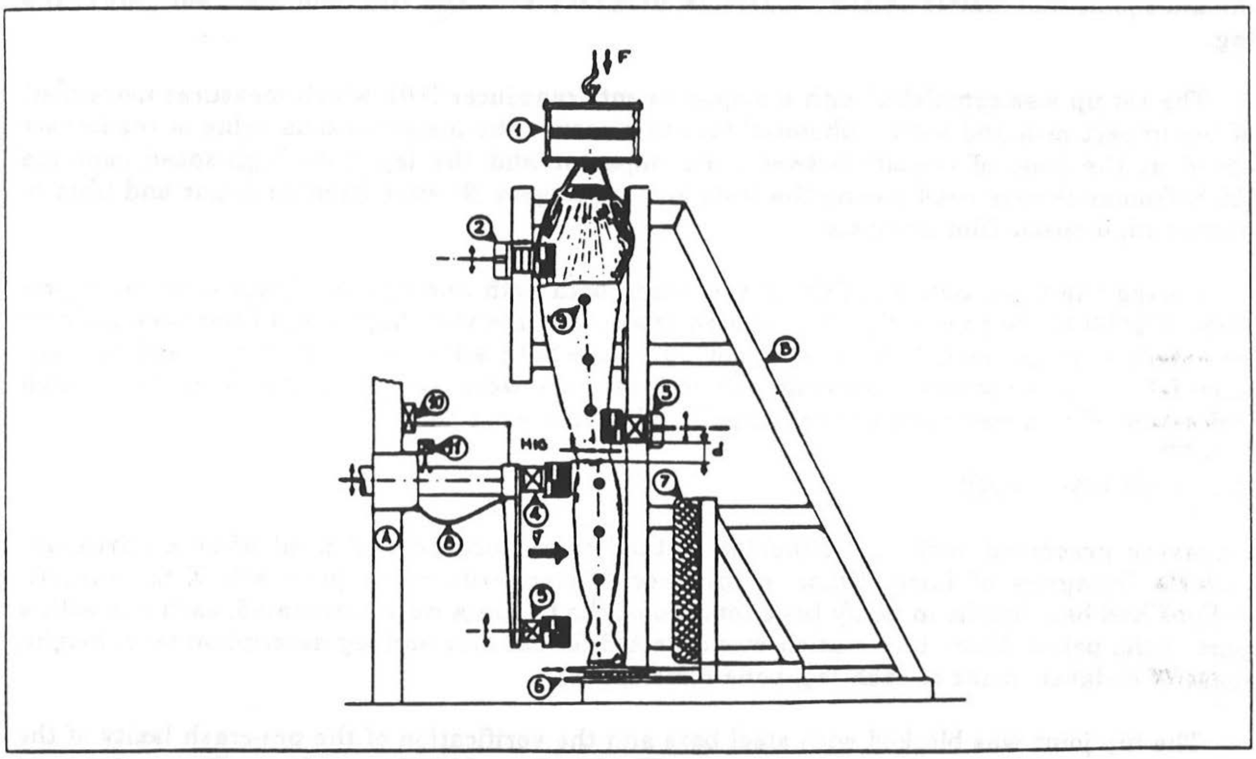

Figure 1. Test set-up.

A - Impactor, B - Support, HIG - Knee joint line, d - Knee free distance,
1 - Pre-loading system,
2 - Upper fixation plate for the leg,
3 - Lower fixation plate for the leg with force transducer,
4 - Upper impact interface (150 mm x 50 mm, 50 mm foam-padded),
5 - Lower impact interface (150 mm x 50 mm, 50 mm foam-padded),
6 - Mobile plate,
7 - Support padding,
8 - Stop wires,
9 - Targets for high-speed cinematography,
10 - Transducer of impactor displacement,
11 - Instantaneous speed measuring cell.

Before the test, the lower limb was positioned and placed on a rigid support (3) and was preloaded with a mass of 40 kg. Two plates, 25 mm foam-padded with Styrodur©, were adjusted for femur stabilisation, one for the proximal part (2) and one for the distal part (3). The lower fixation plate (3) was equipped with one force transducer (SEDEME, 2000 daN) for measurements of the Knee Reaction Force (KRF). This force was also transferred through the knee joint.

Two impact interfaces were mounted on the impactor arm and adjusted to create a translation of the tibia. The upper impact interface (4) was in correspondence with the proximal end of tibia and head of fibula, just below the knee joint. This interface was foam-padded with 50 mm of Styrodur© and equipped with one force transducer (SEDEME, 2000 daN) for measurements of the Knee Impact Force (KIF) and one accelerometer (ENTRAN 250 g). The lower impact interface (5) was adjusted to the level of the distal part of tibia, just above the ankle joint. This interface was foam-padded with 50 mm of Styrodur© and equipped with one force transducer (SEDEME, 2000 daN) for measurements of the Ankle Impact Force (AIF).

Before each test exact positioning of each plate was made to have good orientation of the leg and also to have maximum effect of the shear force on the knee joint. After some preliminary tests the free distance (d), between the lower fixation plate (3) and the upper impact interface (4) was choosen, to be 40 mm, and two impact speed levels (15 km/h and 20 km/h) were used. To permit good movement of the tibia, the foot was placed on a mobile plate (6), this solution minimizes the influence of ground friction at the initial part of the test. Angular movement of the knee joint was limited by the support-padding (7) placed 150 mm from the lower part of the leg.

The set up was completed with a displacement transducer (10), which measures movement of the impact arm and with a photocell (11) to measure the instantaneous value of the impact speed at the time of contact between the impactor and the leg. Two high-speed cameras (1000 frames/s) were used during the tests and six targets (9) were fixed on femur and tibia to permit high-speed film analysis.

During the tests, data acquisition was made with both analog and digital systems. A pre-filter of 1000 Hz was used. Synchronization of electric data with high-speed films was made by an external trigger signal. AD conversion was made with a frequency of 10 KHz and for each channel file 3200 values covering 320 milliseconds were saved on the computer. Batch processing of data was made on the computer with filter class 180.

2.2 MEDICAL PART

Cadavers preserved with the Winckler method were choosen and used after a particular criteria (integrity of lower limbs, corpulence). Afterwards when pre-crash X-ray investigations had been made, to verify bone conditions, the two legs were amputated, each one with a part of the pelvis. Some information was collected for cadaver and leg description (age, height, mass of cadaver, mass of each leg, bone dimensions).

The hip joint was blocked with steel bars and the verification of the pre-crash laxity of the knee joint was made with an arthrometer KT 1000 from MEDMetric Corp. This instrument measures relative displacement between the distal part of the femur and the proximal part of the tibia under loading of the patella-femur complex. We used a force level of 8.9 daN and a knee flexion of 20°. Analysis of differences in knee laxity between left and right leg was made for each cadaver.

After pre-crash laxity measurements, the leg was prepared for mounting under the preload system in the support, using inclusion of a metallic adapter with plaster of paris. After this preparation, the lower limb was mounted on the support (B) as shown on Figure 1 and the preload force was applied. The thigh was blocked with lateral and medial fixation plates: the external one (2) was placed level with greater trochanter, the internal one (3) was adjusted to be level with the medial femoral condyle. Lateral loading of the leg was applied with two impact plates in such a way that forces were applied on a level with tibial lateral condyle and head of fibula by the upper impact interface (4), and level with lateral malleolus by the lower impact interface (5).

After the test, the lower limbs were examined by X-ray. In case the leg had no fractures, post-crash laxity measurements were performed with the same method as during pre-crash.

Thereafter fractures, ligament and meniscus injuries were verified during a dissection of the lower limb.

3. RESULTS

3.1 CADAVER DATA

Eleven cadavers were used in our tests. The main characteristics of those cadavers are shown in Table 1.

Subject #	Sex	Age [years]	Length [m]	Weight [kg]	Livi's index	Relative mass of specimen [%]
LBA 24	F	80	1.54	65	Fat	18.5
LBA25	M	74	1.75	79	Corpulent	20.6
LBA26	M	85	1.67	58	Normal	19.0
LBA 27	M	68	1.75	72	Normal	20.8
LBA 28	M	75	1.77	57	Very thin	19.3
LBA29	F	85	1.56	51	Normal	20.0
LBA30	F	67	1.63	54	Normal	17.6
LBA31	M	67	1.73	93	Fat	17.2
LBA32	F	86	1.56	63	Fat	17.5
LBA 33	F	77	1.60	65	Fat	18.5
LBA34	M	78	1.60	73	Fat	19.2
Mean(±sd)	-	78 (±7)	1.65 (±0.09)	66(±12)	-	18.9 (±1.2)
Min/Max	-	67/87	1.54/1.77	51/93	-	17.2/20.8

Table 1. Test conditions.

The mean value of the cadavers' age was 78 (±7). This value is representative for one of the two groups which are most often involved in car-pedestrian accidents (EEVC/CEVE, 1982).

Livi's index was used to describe different body constitution. Calculation and scale of the index is shown in Figure 2, where M is body mass and T is body height.

Livi's index

$$Li = \frac{10\sqrt[3]{M}}{T}$$

$Li \leq 22$	Subject Very Thin
$22 < Li \leq 23$	Subject Thin
$23 < Li \leq 24$	Subject Normal
$24 < Li \leq 25$	Subject Corpulent
$Li > 25$	Subject Fat

Figure 2. Livi's index, calculation and scale.

Relative mass of the specimen (Table 1) is the ratio of mean mass of left and right legs and the total mass of the cadaver body in percent.

3.2 SPECIMEN DATA

Twenty-one legs were used in the tests. Specimen data is presented in Table 2. The leg mass influences the dynamic response of the specimen during impact. Femur and tibia length, femoral condyle and tibial condyle width are descriptive parameters of the principal form of bones.

Test #	Subject # LBA	Leg mass [kg]	Femur length [mm]	Tibia length [mm]	Femoral cond. width [mm]	Tibial cond. width [mm]	Knee laxity [mm]
FCG06	25 R	16.0	485	370	90	85	4
FCG07	28 R	11.0	478	380	92	85	4
FCG08	28 L	11.0	480	377	89	84	5
FCG09	24 L	12.0	427	320	82	78	6
FCG10	24 R	12.0	421	329	87	78	3
FCG11	26 R	11.0	467	358	85	82	1
FCG12	26 L	10.5	488	354	85	80	1
FCG13	27 L	15.0	488	367	90	88	12
FCG14	27 R	14.5	488	360	90	87	4
FCG15	29 R	10.8	450	378	90	80	1
FCG16	29 L	9.6	447	375	83	80	5
FCG17	30 R	9.8	485	405	95	75	5
FCG18	30 L	9.8	480	410	100	77	8
FCG19	31 L	16.0	510	410	98	88	3
FCG20	31 R	16.0	498	403	95	85	5
FCG 21	33 R	12.0	448	375	85	82	7
FCG 22	32 R	11.0	435	362	77	71	1
FCG 23	34 R	13.0	480	372	91	77	6
FCG24	33 L	12.0	449	365	85	82	9
FCG 25	32 L	11.5	433	361	77	74	1
FCG 26	34 L	14.0	443	370	81	70	3
Mean(±sd)	-	12.3 (±2.1)	464 (±25)	372 (±21)	88 (±6)	80 (±5)	4.5 (±2.9)
Min/Max	-	9.6/16.0	421/510	320/410	77/100	70/87	1/12

Table 2. Specimen data. L - left leg, R - right leg.

The most interesting parameter for description of the knee joint condition is the knee laxity. To determine this value we used Anterior Drawer Measurement in about 20° flexion of the knee joint and with a load of 8.9 daN on the arthrometer (MEDMetric 1982). If the difference of the laxity between the right and left leg was more than 3 mm, the leg with high laxity was discarded from the dynamic test. For this reason the left leg from cadavers LBA 27 and LBA 29 were not used. Later during the medical investigation it was found that both knee joints of these legs presented old injuries without X-ray manifestation.

3.3 DYNAMIC TESTS

Nineteen tests were performed, nine at a velocity of 15 km/h and ten at 20 km/h.

To perform analysis of every test, the information about the kinematics of the leg from high-speed films and the dynamic response measured with the force transducers was used.

Analyses of the high-speed films showed that all injuries occur during the first 30 milliseconds after impact. For this reason, we used only force values within the time window 0 - 30 milliseconds after impact (Figure 3).

When the maximum force occurs outside the injury window, this force has no correlation with the primary injuries, but with a resultant rearrangement of the injured leg.

Analysis of the kinematic and dynamic responses enables the determination for each injury of the time when it occurs and values of the forces at this time. During these investigations two successive time windows were considered, where one can see: for the first one local dominance of the impact force (Knee Impact Force) and for the second the effect of the force transferred through the knee joint (Knee Reaction Force).

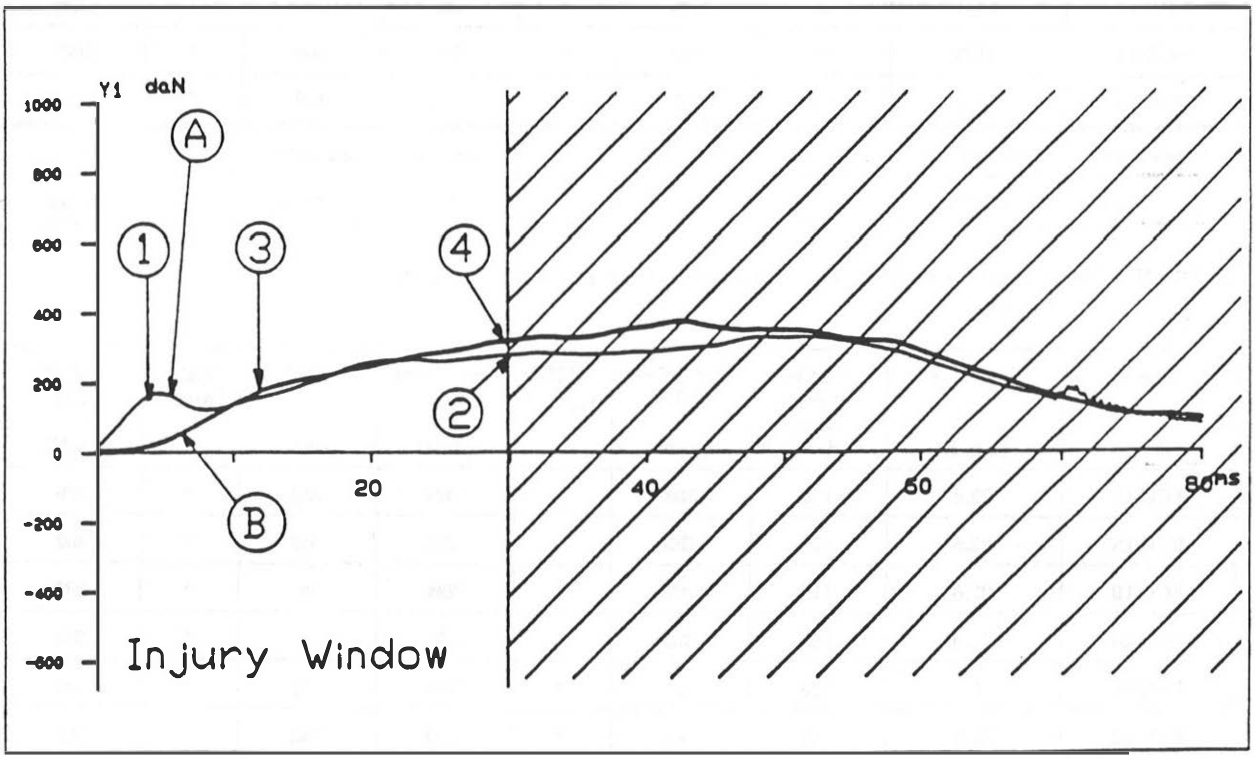

Figure 3. Typical characteristics of Knėe Impact Force (A) and Knee Reaction Force (B).

Typical characteristics of each force are shown in Figure 3, where some particular points can be mentioned:

Point 1: First top value of Knee Impact Force with time correspondence of local injury around the impact point (1 TV KIF). This phenomenon occurs about 5 milli-seconds after impact.

Point 2: Maximum value of Knee Impact Force during the injury window (TV KIF).

Point 3: Value of Knee Reaction Force, at the time when destabilisation of the knee joint occurs (KRF).

Point 4: Maximum value of Knee Reaction Force in the injury window (TV KRF).

The test results at a velocity of 15 km/h are shown in Table 3 and the results at a velocity of 20 km/h in Table 4. KIF and KRF injury indicate only fractures, ligament and meniscus injuries which were found during dissection of the lower limb.

Test #	Velocity [km/h]	Ankle impact force [daN]	1st TV of KIF [daN]	KIF injury	2nd TV of KIF [daN]	KRF [daN]	KRF injury	TV of KRF [daN]
FCG 06	14.2	74	193		287	268		289
FCG07	15.7	74	212		325	276	*	344
FCG 08	14.8	106	165		218	221	*	251
FCG09	14.7	59	156	*	252	241	*	283
FCG10	15.2	123	167	*	241	268	*	293
FCG11	15.8	113	169	*	279	282	*	320
FCG 12	14.8	21	179	*	282	268	*	319
FCG14	15.6	50	254	*	321	305	*	360
FCG 15	15.7	131	121	*	213	182	*	240
Mean(±sd)	15.2 (±0.6)	83 (±37)	160 (±38)	-	268 (±40)	257 (±37)	-	300 (±40)
Min/Max	14.2/15.8	21/131	121/254	-	213/325	182/305	-	240/360

Table 3. Test results of dynamic response. Test series of 15 km/h.
** - injuries.*

Test #	Velocity [km/h]	Ankle impact force [daN]	1st TV of KIF [daN]	KIF injury	2nd TV of KIF [daN]	KRF [daN]	KRF injury	TV of KRF [daN]
FCG 17	22.6	108	292	*	382	350	*	379
FCG18	22.5	107	253	*	383	346	*	402
FCG 19	21.6	119	287	*	234	x	*	397
FCG 20	21.8	109	328	*	320	x	*	383
FCG 21	††	156	264	*	308	328	*	386
FCG 22	21.1	163	187	*	230	243	*	309
FCG 23	20.3	143	222	*	287	315		381
FCG 24	††	122	199	*	219	x		288
FCG 25	††	109	247	*	††	292	*	314
FCG 26	21.2	132	289	*	327	386		410
Mean(±sd)	21.6 (±0.8)	127 (±21)	257 (±45)	-	297 (±59)	322 (±46)	-	365 (±44)
Min/Max	20.3/22.6	107/163	187/328	-	219/382	243/385	-	288/410

Table 4. Test results of dynamic response. Test series of 20 km/h.
*†† - measurement failed, x - value not calculated, * - injuries.*

3.4 INJURY DESCRIPTION

All injuries are presented in Table 5.

Test #	Femur			Tibia			Head of fibula	Ligaments injury				Knee laxity
	diaph. fr.	cond. fr.	cart. inj.	emin. fr.	cond. fr.	diaph. fr.	fr.	LCL	MCL	ACL	PCL	incr. [mm]
FCG06												3
FCG07			•	•					•			2
FCG08			•	•						•		2
FCG09					•		•	•		•		†
FCG10		•		•			•	•		•		†
FCG11			•	•			•	•		•		13
FCG12			•	•			•	•		•		11
FCG14				•			•	•		•		9
FCG15		•			•	•	•	•		•		†
FCG17	•						•			•		†
FCG18					•		•			•		†
FCG19			•	•	•		•			•		†
FCG20		•	•	•	•		•		•	•		†
FCG21	•			•			•			•		†
FCG22							•		•	•		4
FCG23						•	•					†
FCG24							•	•				††
FCG25				•			•			•		7
FCG26							•	•				2

Table 5. Injuries.
• - injury observed, † - fractures, no measurements, †† - measurement failed.

In Table 5 a presentation of injury occurrence is made, without any scaling of severity. Many different kinds of injury are grouped under the same column title:

1. Fracture of the femoral diaphysis include simple or transverse fractures, simple oblique fractures.
2. Fracture of the femoral condyles include comminute or simple, intra- and extra-articular fractures.
3. Injury of the femoral cartilage is a typical consequence of relatively displacement between femur and tibia and it occurs at the contact point between tibial intercondylar eminence and medial femoral condyle.
4. Fracture of the tibial intercondylar eminence is typically injury associated with injury described in the point 3.
5. Fracture of the tibial condyles include simple or comminute, intra- and extra-articular fractures with or without compression.
6. Fracture of the tibial diaphysis include simple fractures.
7. Fracture of the head of fibula include simple or comminute fractures with or without compression and extraarticular fracture of neck of fibula.

8. Ligament injury for fibular collateral ligament (LCL), tibial collateral ligament (MCL), anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL) include partial or total ruptur and avulsion of the femoral or tibial attachment.

Increment of the knee laxity was calculated as the difference between the value for anterior drawer after the test and corresponding value before the test in cases when condyle fractures were not found during X-ray investigations. Increment of the knee laxity higher or equal 2 mm corresponds in In Vitro study (MEDMetric 1982) with injury of the anterior cruciate ligament.

4. DISCUSSION

The purpose of this study was to determine the force levels at which injuries to the knee occurs at lateral shearing.

The high-speed film analysis related to the time history of the forces shows the necessity of discussing two different injury mechanisms.

The first injury mechanism under the test conditions of this study is directly related to the Knee Impact Force. Generation of this force is correlated with the local acceleration of the biological system (bone and muscles) and is a form of contact force. The consequences of this force are two different kinds of injury (Tables 3, 4 and 5):

- contact injuries, for example head of fibula and lateral tibial condyle fractures,
- extra-articular injuries and diaphysis fracture of tibia or femur.

This injury mechanism occurs at the beginning of the lateral displacement, for impact velocities 15 - 20 km/h about 5 milliseconds after the impact. The mean peak force value (1st TV of KIF) correlated with injuries described above was (Table 3 and 4):

- 180 (±38) daN for an impact velocity of 15 km/h,
- 257 (±45) daN for an impact velocity of 20 km/h.

The second injury mechanism is correlated with the force transferred through the knee joint during the impact when the thigh is accelerated. This injury mechanism occurs later in the injury window, for impact velocities 15 - 20 km/h about 15 - 20 milliseconds after the impact. Force transferred through the knee joint was, in our test set-up, relative to the Knee Reaction Force. The consequences of this force was relative displacement of tibia and femur and this generates intra-articular injuries of the knee, for example anterior cruciate ligament (ACL), fibular collateral ligament (LCL) or tibial collateral ligament (MCL) ruptur (avulsion) and fracture of tibial intercondylar eminence. Another typical consequence of this relative displacement is femoral cartilage injury which occurs in the contact point between the tibial intercondylar eminence and the medial femoral condyle. Mean peak force value (KRF) correlated with injuries described above was (Table 3 and 4):

- 257 (±37) daN at impact velocity of 15 km/h,
- 322 (±46) daN at impact velocity of 20 km/h.

The difference between the force values for these two velocities is an example of the response from a viscoelastic material. Bone and muscles represent this kind of material. The mechanical characteristics of viscoelastic materials are related to the loading rate.

In our test condition it is necessary to use both parameters, Knee Impact Force and Knee Reaction Force, to describe consequences of a lateral impact near the knee joint. Maximum values of those force signals are not sufficient to represent the injury risk. In Table 3 and 4 values in column "2nd TV of KIF" and "TV of KRF" are generally higher then corresponding force values which generate injury in the knee region. These maximum values of forces have no correlation with the primary injuries, but with a rearrangement of the injured leg.

To describe consequences of a lateral impact to the knee joint and tolerance to shear loads, the impact force (in our test condition Knee Impact Force) is not adequate, because this force primarily is related to local injuries at the impact point.

In our test condition the impact level was choosen to be near the knee joint for maximum effect of the shear loads just at the joint. The Knee Impact Force level (1st TV of KIF) in this impact configuration describes realistically only ultimate strength of the head of fibula and the tibial lateral condyle for direct loading. In pedestrian accidents it has correspondence with a configuration where the bumper hits the leg from the side just below the knee joint.

Injuries inside the knee joint are generated by the force transferred through this joint (Knee Reaction Force). At this type of loading the most frequent injury found during the dissection was avulsion or rupture of the anterior cruciate ligament - in 14 tests of 19 (Table 5) and fracture of the tibial intercondylar eminence - in 10 tests. Also the lateral collateral ligament was injured in 8 tests. No injuries of the posterior cruciate ligament were observed.

The typical injury at the impact area was a fracture of the head or neck of the fibula. In 16 tests this kind of fractures were observed but only in 5 tests fractures of the tibial lateral condyle were found. It seems that the strength of the head of fibula is lower then that of the tibial lateral condyle. This first mentioned structure has some damping effect at direct impact and this way protects the tibial lateral condyle. Injuries to the fibula were generated in each test at velocity of 20 km/h.

Another indicator of knee injuries, foremost the anterior cruciate ligament, were measurements of the knee laxity. Investigation of the knee laxity before each test (Table 2) gave important information about the conditions of the two knee joints from the same cadaver. This kind of testing is necessary to find effects from old injuries to the knee joint and to make a selection of legs before the test. Calculation of knee laxity increments (Table 5) shows that even in test FCG 06 the knee joint had minor injuries (ligament stretching) without manifestation during the dissection. In pedestrian impact simulation with cadavers these measurements are very useful for detection of minor injuries at the knee joint.

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