Response of the Knee Joint in Lateral Impact: Effect of Bending Moment

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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. This dummy enables measurement of biomechanical parameters, such as forces and moments at the knee joint level related to the injury mechanisms. To determine the resistance of the human knee to shear force or bending moment and to describe the injury mechanisms, it was necessary to make separate *In-Vitro* experiments with human cadaver specimens, where only one of those two parameters affects the biological material at the time.

In the present study an experimental method for assessing the tolerance to bending moment in the lateral direction of the extended knee joint has been developed. The load response of the lower extremity was measured by means of force transducers. The bending moment transferred through the entire knee joint was calculated and the motion of the specimen was registered by high speed photography. The bending moment in the lateral direction at the first sign of damage of the entire extended knee joint was determined. Damages were assessed by measuring the knee joint condition (valgus-varus and anterior drawer increment) and by dissecting the knee region. Seventeen tests were carried out under dynamic conditions, seven at a velocity of 16 km/h and ten at 20 km/h on human cadaver lower extremities.

The first and most common damage type to entire knee joint in this loading configuration was stretching and rupture or avulsion of the medial collateral ligament (MCL). These damages are generated by dentical mechanisms i.e. tensional forces acting on the medial knee structures. The mean peak moment correlated with this damage mechanism was 101 (±21) Nm for an impact velocity of 16 km/h, and 123 (±35) Nm for an impact velocity of 20 km/h. The mechanisms led to damage of the knee joint when the lower extremity was bent approximately 10° in the lateral direction at the knee joint.

INTRODUCTION

Pedestrians form a major group of traffic accident victims. A statistical investigation of traffic accidents in some European countries shows that approximately 13 - 33 % of all road accident death were pedestrians (EEVC/ CEVE, 1982). It is notable though that the percentage of pedestrians killed in all road accidents is higher in less motorised countries than in highly motorised countries. Malini and Victor (1990) observed that as many as 35 - 45 % of all road accident casualties were pedestrians in the less motorised countries, and these accidents were particularly common in the urban areas.

Pedestrians often cross a street in a more or less perpendicular direction when they are hit from the side by the front structure of a car. These pedestrians are usually elderly people and therefore more fragile than younger ones. The first contact occurs between the most protruding structure of the car front, which is usually the front bumper, and the lower extremity of the pedestrian (Appel et al., 1975; Lestrelin et al., 1985).

Since this kind of accident usually occurs in built-up areas, the speed of the car in most cases is quite low, generally below 30 km/h (Tharp and Tsongos, 1976; Danner et al., 1979). At higher impact-speed head injuries is the most serious outcome. At lower speeds it is knee injuries that are the main concern (Otte, 1989) because they often require long periods of treatment. Follow-up studies have shown high frequencies of long term impairment for this category.

The knee joint can be regarded as the biomechanical link between the thigh and the leg. When impact occurs close to the knee joint, it generally leads to forces that turn the leg outward around an assumed axis corresponding to the course of the impacting car front. The first consequence of these forces are injuries in the contact area as well as extra-articular injuries. Due to the forces transmitted through the knee joint, a transverse dislocation between the leg and the thigh occurs. This leads to a luxation of the knee joint accompanied by the over-stressing and over-stretching of all affected parts of the ligamentous structures. The consequence of this luxation is intra-articular injuries of the knee joint.

Another obvious effect of the impact of the car bumper on the lateral aspect of the knee is that the lower extremities bend laterally. This lateral bending results in tensile forces on the medial side of the knee and compressive forces in the lateral condyles. Usually the medial collateral ligament is stretched and may be torn from one of its attachment areas. These forces often also result in compression fractures of the lateral condyle of the tibia. These fractures usually extend into the knee joint. The inertia of the foot causes the leg to rotate about its longitudinal axis, resulting in shear in the soft tissues at contact with the bumper and possibly also torque which is transmitted to the knee region. The worst case in car-pedestrian accidents seems to be when the bumper impacts on the lower extremity that supports the body and when this impact occurs close to the extended knee joint (Kajzer, 1989).

During the 80's a new type of crash impact dummy, the rotationally symmetrical pedestrian dummy (RSPD), suitable for assessment of car front aggressiveness in pedestrian impacts was developed (Aldman, 1985). This dummy enables measurements of biomechanical parameters, such as forces and moments at the knee joint level, that are directly related to the injury mechanisms. The development of the RSPD has made it possible to assess shear force and bending moment separately.

A standardised test method for the evaluation of car-front aggressiveness to pedestrian lower extremities in a car-pedestrian collision is being promoted by the EEVC (Harris, 1989 and 1991), the NHTSA (Elias and Monk, 1989; Kessler and Monk, 1989) and the ISO. Currently, the ISO/TC22/SC10/WG2 (Pedestrian Impact Test Procedure) is considering procedures for testing car-front aggressiveness to pedestrian lower extremities.

The setting up of standardised subsystem test methods for the assessment of car-front aggressiveness to pedestrian lower extremities requires certain data about the tolerance to lateral impact of the human lower extremities, especially of the knee joint. These data are important in terms of the tolerance level to the shear force and the bending moment. Unfortunately, such data are lacking.

For this reason the Department of Injury Prevention at Chalmers University of Technology in Göteborg, Sweden and two laboratories of Institut National de Recherche sur les Transports et leur Securite (INRETS) in France: the Laboratory of Chock and Biomechanics (LCB) in Bron and the Laboratory of Applied Biomechanics (LBA) of the Faculty of Medicine in Marseille, decided to carry out special studies aimed out describing the influence of shearing and bending loads on the entire knee joint and determining the resistance to shear force and bending moment of this joint. These experimental studies were made at the LBA.

The first phase of the project was a description of the effects of shear forces when applied to the human knee region in the lateral direction. Results from this phase of the project have been presented and discussed in detail by Kajzer et al. (1990). An investigation of the effects of the bending moment in the lateral direction on the human knee region was the subject of the present study and makes up the second phase of the cooperation project.

METHODOLOGY

Nine cadavers preserved by means of the Winckler method were choosen according to a particular criterion that of integrity of lower extremities and corpulence. After pre-crash X-ray investigations had been made to verify bone conditions and to exclude pathological changes, the two lower extremities were amputated, each with a part of the pelvis attached to it.

The following demographic and morphometric data were collected for the description of the cadavers and lower extremities: sex, age, length, mass of cadaver, mass of each lower limb, femur and tibia dimensions.

Femur and tibia lengths, femoral condyle and tibial condyle widths were measured by X-ray photographs. Livi's index was used to describe the body constitution of the subjects. The calculation and the scale of the index are shown in Figure 1.

Livi's Index

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

$Li \leq 22$	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 1. Livi's index, calculation and scale. M - body mass in kilograms; H - body height in meters.

The relative mass of the specimen was calculated as the ratio of the mean mass of the left and right legs and the total mass of the cadaver in percent. This parameter was used to describe the mass distribution between different body segments and as a complement to Livi's index to indicate the corpulence of the lower extremities.

The knee joint condition was assessed by means of the valgus-varus and knee laxity measurements. The pre-crash valgus-varus was measured when the distal part of the leg was loaded with 30 N. The pre-crash laxity of the knee joint was assessed with an arthrometer KT 1000 from MEDMetric Corp. This instrument measures the relative displacement between the distal part of the femur and the proximal part of the tibia while the patella-femur complex is being loaded. To measure pre-crash laxity we used the anterior drawer measurement at about 20° flexion of the knee joint and with a load of 8.9 daN on the arthrometer (MEDMetric 1982).

An analysis of the differences in pre-crash knee laxity between left and right lower extremity was made for each cadaver. If the difference in laxity between the right and left lower extremity was more than 3 mm, the extremity with high laxity was eliminated from the dynamic test, because this could indicate that there is a risk of old injuries to knee joint.

After pre-crash valgus-varus and laxity measurements, the hip joint was blocked by steel bars and the lower extremity was mounted under the preload system in the support by means of the inclusion of a metallic adapter with plaster of paris. A total of seventeen lower extremities were accepted for dynamic testing.

Dynamic testing was done by means of an impactor propulsed by a rubber band. The mass of the mobile part of the impactor was approximately 40 kg. A specially designed impactor face equipped with a force transducer was mounted in front of the impactor. Initial test conditions used in the present study are shown in Figure 2.

The experimental set-up was designed so that a dynamic flexion load that would cause a lateral flexion type of damages to the knee region would be applied. Before each test the lower extremity was placed on a rigid support (B). It was positioned such that it could be extended and preloaded with a mass (1) of 40 kg, so that maximum effect of the bending moment at the knee joint, and simulation of the worst case of car-pedestrian accident could be achieved.

The thigh was blocked with upper and lower fixation plates: the internal plate (2) was placed level with pubic bone, and the external plate (3) was adjusted to be level with the lateral femoral condyle. The free distance (d), between the lower fixation plate (3) and the knee joint line was chosen, to be about 20 mm. These plates consisted of foam-padded with 25 mm of Styrodur© and were equipped with two force transducers (SEDEME, 20 kN). The load response of the knee joint (upper reaction force and the lower reaction force) was measured on the upper fixation plate and the lower fixation plate by means of these two transducers. The bending moment, transmitted through the entire knee joint (knee bending moment), was calculated by means of the upper reaction force and the lower reaction force and the corresponding distances to the knee joint line (HIG).

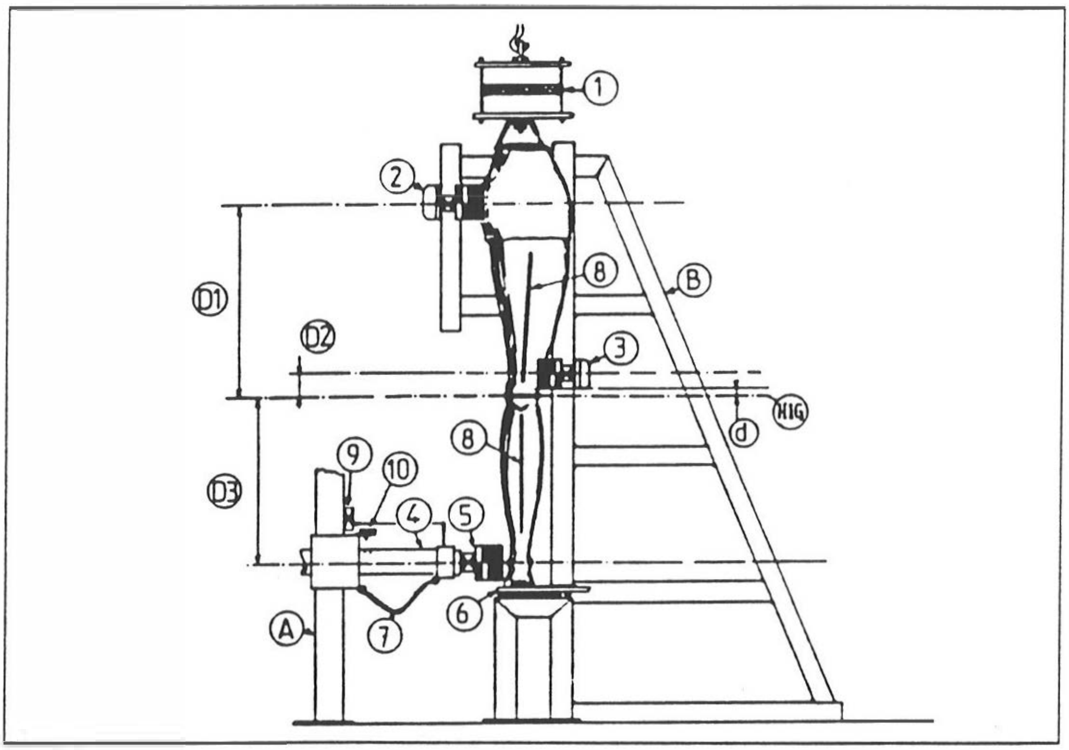

Figure 2. Experimental test set-up for dynamic loading. A - Impactor; B - Support; HIG - Knee joint line; d - Knee free distance; 1 - Pre-loading system; 2 - Upper fixation plate for the thigh with force transducer; 3 - Lower fixation plate for the thigh with force transducer; 4 - Impactor; 5 - Impactor face with force transducer and accelerometer; 6 - Mobile plate; 7 - Stop wires; 8 - Lines for high-speed cinematography; 9 - Transducer of impactor displacement; 10 - Instantaneous speed measuring cell.

A dynamic lateral load was applied to the leg by one impactor face. This impactor face (5) was mounted on the impactor (4) and adjusted to create bending of the lower extremity at the knee joint similar to that which occurs when a pedestrian is impacted from the side by the car-front. The impactor face was adjusted to the level of the distal part of the tibia, just above the ankle joint so that impact occurred on the medial side of the leg. This face was 150 mm x 50 mm, foam-padded with 50 mm of Styrodur© and equipped with one accelerometer (Entran EGA, 250 g) and one force transducer (SEDEME, 20 kN) for measurements of the ankle impact force. A signal from this accelerometer was used to describe the load pulse applied to the leg. The free movement of the impactor was limited to about 250 mm (150 mm in tests MFG P01 and MFG P02) by means of four stop wires (7).

To permit good movement of the tibia, the foot was placed on a mobile plate (6). This solution minimised the influence of ground friction in the initial stage of the test.

The set-up was completed with a displacement transducer (9), which measures the movement of the impactor, and with a photocell (10) which measures the instantaneous value of the impact speed at the time of contact between the impactor face and the leg.

The motion behaviour of the lower extremity was monitored by two high-speed cameras (Stalex WS2 and WS3, 1000 frames/s). Two lines (8) were drawn on the thigh and the leg to permit high-speed film analysis of the lateral bending angle at the knee joint. This lateral bending angle was calculated as the difference between the position of these lines before the test and position after impact. The lateral bending angle was calculated during the whole loading sequence and correlated to the load response of the knee joint. From the test number MFG 12, one additional high-speed camera (Stalex WS2, 1000 frames/s) was used from the side to detect rotation of the leg around longitudinal axis.

During the tests, data was registered by a digital system (Techniphone-Mors BA32) connected to a Hewlett-Packard computer. A pre-filter of 1000 Hz was used. Synchronisation of electric data with high-speed films was done by an external trigger signal. AD conversion was achieved at a frequency of 10 kHz. For each channel file 3200 values covering 320 ms were saved and batch processing of data with filter class 180 was done on the HP computer.

The shape of the load pulse (magnitude, rise time, duration) that was applied to the leg was a result of the velocity of the impactor and the force-deformation characteristic of the padding material attached to the impactor face. The two load pulses were achieved with two speed levels of the impactor (16 km/h and 20 km/h) as presented in Table 1. The difference between two load pulses was in the magnitude of the acceleration. The rise time and the duration of the acceleration were very similar for both velocities of the impactor. From this combination of the three parameters, comparisons could be made in order to investigate whether the load response of the entire knee joint was sensitive to the loading rate.

Load pulse (velocity of the impactor)	16 km/h	20 km/h
Peak acceleration (g)	7.0 (±0.4)	9.8 (±2.0)
Rise time of acceleration (ms)	3.6 (±0.2)	3.6 (±0.2)
Duration of acceleration (ms)	8.2 (±0.7)	8.4 (±0.7)

Table 1. Description of the dynamic load pulses used in the experiments. Mean (±SD) values are presented.

After the test, the lower extremities were X-rayed. Thereafter, post-crash valgus-varus and knee laxity measurements were performed with the same method as during the pre-crash investigation. Fractures and ligament damages were verified and assessed by a dissecting the lower extremities.

The first sign of damage of the entire knee joint was established by valgus-varus or knee laxity measurements and then defined as a distinct increment of the knee instability or as partial or total failure of knee joint structures. The first sign of damage was detected by the changes in the knee bending moment curves in the form of a distinct dip.

The increment of the knee laxity was calculated as the difference between the value of the anterior drawer after the test and the corresponding value before the test. As knee-damage criterion, the increment of the knee laxity higher or equal 2 mm was used, which corresponds in the *In-Vitro* study (MEDMetric 1982) to damage of the anterior cruciate ligament.

The increment of the valgus-varus was calculated as the difference between values for the valgus-varus after the test and corresponding values before the test. No criterion for knee damage corresponding to this measurement was available. It is well known that increment in valgus-varus strongly correlates with the damage of the medial collateral ligament and the lateral collateral ligament.

MATERIAL

Nine male cadaver specimens were available for this study. The mean age was 77 (±9) years which is representative for one of the two groups most often involved in car-pedestrian accidents (EEVC/CEVE, 1982). The length and the weight of cadavers show a normal variation for the population in France. Subjects with varying body constitution, described by Livi's index as very thin, normal, corpulent or fat, were represented. The relative mass of the specimens shows a variation from 17.6% to 25.0% with a mean value of 20.9 (±2.2)%, one cadaver (LBA 49) deviates in terms of this value.

Eighteen lower extremities were amputated from the available cadavers. Seventeen lower extremity specimens were used in dynamic tests. One lower extremity was eliminated from the dynamic test due to problems with conservation.

The value of the mass of each lower extremity showed high variation between the different cadavers, from 12 kg to 17 kg. Especially for LBA 49, described according to Livi's index as normal, the value of the mass was high, 16 kg and 15 kg for the left and right leg, respectively. Differences in mass between the left and right lower extremity of the same cadaver were small. The dynamic response of the leg during impact is influenced by the mass of the leg that is in relation to the mass of the lower extremity.

The main biometry of the bones showed that cadavers used in the present study represented the normal variation for adults.

The pre-crash knee laxity (normal anterior drawer) showed a variation of 0 - 9 mm with a mean value of 2.8 (±3.0) mm. A study presenting variations of values of normal anterior drawer for cadavers preserved with the Winckler method was not available for comparison. In one study with fresh cadaver specimens, a mean value of 5.9 mm was noted for this parameter (MEDMetric, 1982). The difference in knee laxity between two legs from the same cadaver was for all cases lower or equal to 3 mm. Corresponding to the selection criterion used all specimens were accepted for dynamic tests.

RESULTS

Seventeen tests were performed: seven at a velocity of 16 km/h and ten at 20 km/h. The results from the medical investigation, the dynamic response measured with the force transducers and the information about the kinematics of the lower extremities achieved with high-speed films were used in the analysis of each test.

Damage Description

All damages found in the medical investigation are presented in Table 2. In this table the calculated values for knee laxity increment and valgus-varus increment are also presented.

Test #	Femur			Tibia			Joint Capsule Damage	Ligaments Damage				Knee laxity/ Valgus-Varus
	Diaph. Fr.	Cond. Fr.	Cart. Damage	Emin. Fr.	Cond. Fr.	Diaph. Fr.		LCL	MCL	ACL	PCL	Incr. [mm/]
MFGP01									*			8/4
MFG01									*			6/6
MFG02									*			3/4
MFG03									*			4/3
MFG04								*	*			6/3
MFG05												2/0
MFG06												0/0
MFGP02									*			5/3
MFG07									*	*		3/9
MFG08												1/2
MFG09									*			3/5
MFG 10					*							4/4
MFG 11												0/0
MFG 12							*		*			1/11
MFG13					*				*			1/8
MFG 14									*	*		5/13
MFG 15							*		*	*		12/7

Table 2. Damages (* - indicates observed damage at the medical investigation).

In Table 2, a summary of damage occurrence without any scaling of severity is presented. Many different kinds of damage are grouped under the same column title:

1. Fracture of the femur and the tibia including simple or comminute, intra- and extra-articular fractures.

2. Ligament damage for lateral collateral ligament (LCL), medial collateral ligament (MCL), anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL) including partial or total rupture and avulsion of the femoral or tibial attachment.
3. Joint capsule damage including partial or total rupture.

Table 2 shows clearly that lateral bending loading of the lower extremities results mainly in a knee ligament damage. The most common type of damage was rupture or avulsion of the MCL. Only two tibial condyle fractures inside the joint capsule were found, and these fractures occurred without diaphysis fractures. Rupture or avulsion of the MCL was found in about 70% of the specimens investigated (12 cases of 17). In six specimens other damages were found to be associated with MCL damages: ACL damages in three cases; joint capsule damages in two cases; LCL damage in one case and tibial condyle fracture in one case. Only in MFG 10 was the tibial condyle fracture found as a single damage. In MFG 10 a tibial condyle fracture could occur due to differences in strength of the ligaments and bones of the specimen.

In four cases no damages were found during the medical investigation. Even measurements of knee instability in these four specimens, according to the damage criterion used, did not indicate ligament damages (MFG 05, MFG 06, MFG 08 and MFG 11).

When the entire knee joint was exposed to lateral bending the valgus-varus increment shows a better relation to ligament damages than does the knee laxity increment. The value of the valgus-varus was higher when the ACL damage was associated with the MCL damage (MFG 07) than when MCL damages had occurred alone (MFG P01 and MFG P02). Values noted for the increment of valgus-varus and knee laxity were low, and belong the chosen damage criterion for ACL when there were no knee ligament damages found in the medical examination (MEDMetric 1982).

Dynamic Response of the Knee Joint

Analyses of the high-speed films showed that in the present study the first sign of damage occurs during the first 30 ms after impact. For this reason in the analysis only force and moment values within the time-window 0 - 30 ms after impact were used (Figure 3). When the maximum force or moment values occur outside the damage window, these values have no correspondence with the first sign of damage, but rather with the resultant rearrangement of the damaged specimen.

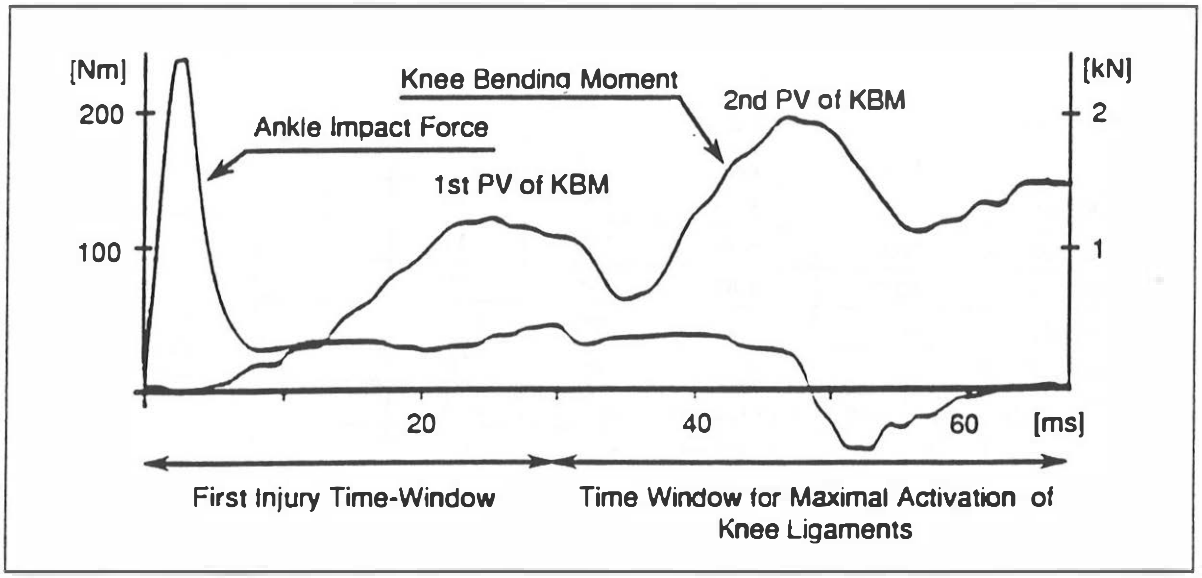

Figure 3. Knee bending moment versus time from a typical test in which the entire knee joint is exposed to a transient load pulse (ankle impact force) at a velocity of 20 km/h.

Analysis of the kinematic and dynamic responses enables the determination of the time of occurrence and the value of bending moment at time for each damage. During these investigations it was found that damages do not occur simultaneously but during two successive time window. The MCL and tibial condyles were the first structures to be damaged during impact on the leg that resulted in the bending deformation of the knee joint.

Typical characteristics of the ankle impact force and the knee bending moment are shown in Figure 3. Some particular points head to be mentioned:

- Peak value of ankle impact force, which in our set-up occurred at 2.7 (±0.3) ms and 2.7 (±0.4) ms after impact with velocities of about 16 km/h and 20 km/h, respectively. It is important to point out that this force has no direct relation to knee damages but was a result of the acceleration of the leg. However its kinematic energy can later generate damages in the knee joint. This force is a product of the active mass of the specimen and its acceleration.
- Bending moment at the first sign of damage defined as the first peak value of the knee bending moment (1st PV of KBM) before the first distinct dip on the moment curve that corresponds to total failure of the MCL or to a condyle fracture. This peak value in our set-up occurred at about 20 (±4) ms after impact.
- The second peak value of the knee bending moment (2nd PV of KBM) before the second distinct dip on the moment curve does not correspond directly to total failure of the ACL. This value represents the maximal activation of all non damaged knee structures, and in our set-up occurred at about 50 (±10) ms after impact. This peak value occurred outside the first damage time-window and had no correspondence with first sign of damage to the knee joint but with a resultant rearrangement of the damaged joint. As this value in not within the scope of this report it was not included in the analysis of test results.

It is obvious that ligament damages in the form of stretching are initiated before total or partial failure occurs. In the present study only the correlation between failure of ligaments or fractures of bones and knee bending moment were studied. The stretching of the ligaments could be determined by measurements of the knee stability, but could not be related to the occurrence of the knee bending moment due to not distinguish change on its time-curve.

The primary goal of this project was, as in the previous study (Kajzer et al., 1990), to perform tests involving two velocities 15 km/h and 20 km/h, and to find the tolerance level of the entire knee joint to lateral bending. Because of some modifications in the impactor machine this low velocity was 16.3 (±0.2) km/h and the higher one 20.2 (±0.3) km/h with a small variation of 19.7 km/h to 20.7 km/h.

Test #	Velocity [km/h]	Ankle Impact Force [kN]	1st PV of KBM [Nm]	Knee Lateral Bending Angle [°]	MCL Damage	Longit. Leg Rotation	Condyle Fracture
MFG P01	16.4	1.54	85	7	*		
MFG 01	16.3	2.08	91	7	*		
MFG 02	16.2	1.85	106	10	*		
MFG 03	16.3	1.70	135	11	*		
MFG 04	16.2	1.61	88	10	*		
MFG 05	16.6	1.73	*(33)*	*(10)*		**	
MFG 06	16.5	1.67	*(48)*	*(11)*		**	
Mean (±sd)	16.3 (±0.2)	1.74 (±0.18)	101 (±21)	9 (±2)	-	-	-
Min/Max.	16.2/16.6	1.54/2.08	85/135	7/11	-	-	-

Table 3. Test results of dynamic response. Test series at 16 km/h (values printed in cursive style were not used in calculation of Mean, Min/Max.; * - damages, ** - rotation).

The test results at 16 km/h are shown in Table 3 and the results at 20 km/h in Table 4. The MCL damage column shows ligament damages detected by measurements on knee instability and by damages found during the medical investigation of the knee joint. The values of the knee lateral bending angle have been calculated at the time when the 1st PV of KBM occurred.

The calculation of the mean, minimum and maximum values of the 1st PV of KBM and the lateral bending angle of the knee was made only for cases when MCL damages or fractures of medial condyles were found in the medical investigations. It was found that these damages were generated by the same mechanisms. The 1st PV of KBM from the test MFG 13 — where the fracture of lateral condyle was observed — has been excluded from the calculations due to different damage mechanisms. This fracture was a result of local bone compression at the contact area of femoral and tibial condyles. A compression fracture was observed only in one test and no statistical calculations could thus be made. It is interesting to point out that the fracture was found in the specimen which knee was exposed to low bending moment. The 1st PV of KBM related to this fracture was only 59 Nm.

Test #	Velocity [km/h]	Ankle Impact Force [kN]	1st PV of KBM [Nm]	Knee Lateral Bending Angle [°]	MCL Damage	Longit. Leg Rotation	Condyle Fracture
MFG P02	20.6	2.06	90	13	*		
MFG 07	20.0	3.02	190	6	*		
MFG 08	19.7	2.57	*(12)*	*(10)*		**	
MFG 09	20.7	2.85	107	11	*		
MFG 10	19.9	3.29	138	12			***m
MFG 11	20.3	1.84	*(63)*	*(15)*		**	
MFG 12	20.0	2.32	137	10	*		
MFG 13	20.0	2.28	*(59)*	*(10)*	*		***l
MFG 14	20.3	1.94	111	14	*		
MFG 15	20.1	1.90	90	11	*		
Mean (±sd)	20.2 (±0.3)	2.41 (±0.51)	123 (±35)	11 (±3)	-	-	-
Min/Max.	19.7/20.7	1.84/3.29	90/190	6/14	-	-	-

Table 4. Test results of dynamic response. Test series at 20 km/h (values printed in cursive style were not used in calculation of Mean, Min/Max.; * - damages, **- rotation, ***m - fracture of the medial tibial condyle, ***l - fracture of the lateral tibial condyle).

A phenomenon which was not described or discussed in previous investigations concerning damage mechanisms of the pedestrian lower extremity has been observed in the present study during the lateral bending-loading of the knee joint. This was a longitudinal leg rotation that resulted in knee flexion. Because the goal of the present study was to determine the tolerance level of entire knee joint for first sign of damage at pure lateral bending the knee joint was extended and the force was applied from the side in such a way that the risk for rotation was minimised. Even in this set-up the leg has a tendency to rotate around its longitudinal axis. In four tests this rotation was observed already in the first damage time-window but it couldnot to be quantitative determined in the set-up used in the present study. This phenomenon resulted in a different activation of the knee structure than that assumed as a goal for present investigation. No damages to the knee joints were observed in tests with longitudinal leg rotation and the 1st PV of KBM calculated in these tests was low.

DISCUSSION AND CONCLUSIONS

The purpose of the present study was to describe the knee joint response to dynamic bending loads and thus determine the moment level at which the first sign of damage to the knee occurs at lateral bending of the extended lower extremity. The high-speed film analysis in relation to the time history of the forces shows that different damage mechanisms, which are activated in different time windows need to be studied. Knowledge of these damage mechanisms is essential for the design and validation of mechanical and mathematical models of the lower extremity. Better knowledge also creates a basis for developing new systems for reducing risk and minimising the severity of injuries, particularly those caused by car-pedestrian accidents.

In the present study the impact level was chosen to be near the ankle joint and at the medial side of the leg. The experimental set-up was designed to simulate maximum effect of the bending moment at the knee joint and the same bending orientation of the knee joint as in real car-pedestrian accidents. In car-pedestrian accidents this load application corresponds to a configuration where the bumper hits the leg from the side just below the knee joint. The bending orientation corresponds to a configuration where the bumper hits the leg from the side between the knee joint and the center of percussion of the leg situated close to its center of gravity.

Damages inside the knee joint are generated by forces transmitted through its ligament structures. The damage mechanism tested in the present study, which generated the first sign of damage and was made active in the first damage time-window, is directly related to the force acting on the MCL. The consequences of this force are three different types of damage (Tables 2, 3 and 4):

- MCL partial or total rupture;
- MCL avulsion;
- fracture of medial condyle.

In this type of loading of the entire knee joint, the most frequent damage found in the medical investigation was avulsion or rupture of the MCL. Fracture of the medial condyle can also occur due to differences in strength between ligament and bone. In medical investigation this kind of damage can be classified as MCL avulsion. During the first damage time-window also ACL, PCL, LCL and joint capsule are exposed to stretching, but it seems that these structures are loaded below their tolerance levels and the loading does not result in damages at this time-window.

The damage mechanism that generated the first sign of damage is activated at the beginning of the lateral bending, for impact velocities of 16 - 20 km/h about 20 ms after impact. The mean peak value of the knee bending moment (1st PV of KBM) correlated with the damages described above was (Table 3 and 4):

- 101 (±21) Nm for an impact velocity of 16 km/h;
- 123 (±35) Nm for an impact velocity of 20 km/h.

The results indicated that the load response of the knee joint was higher when the lower extremity was exposed to a more rapid load. The difference between the two data sets from tests performed at two velocities used in the present study was not statistically significant on the 5% level. The difference between the force values of these two velocities is an example of response from a viscoelastic material. The mechanical characteristics of viscoelastic materials are influenced by the loading rate. The ligament tissue is a viscoelastic material and is strain-rate sensitive. The ligament increases in strength and stiffness with increased speed of loading.

The outcome of an *In-Vitro* damage is dependent not only on the magnitude and direction of the load, but also on inherent properties of the damaged structure itself. The coefficient of correlation between the knee bending moment (1st PV of KBM) and the biological parameters of specimens that were collected in the present study in terms of sex, age, length, weight, femur length, tibia length, femoral condyle width, tibial condyle width and knee laxity were determined for both test series. The present study shows that results did not result in a significant correlation between biomechanical properties of the entire knee joint and the demographic and morphometric data collected for cadavers and lower extremity specimens. A significant correlation was only found for the bending moment at the first sign of damage (1st PV of KBM) and the tibial condyle width, the coefficient of correlation being $r = 0.95$ and $r = 0.65$ ($p \leq 0.05$) in test series with impact velocities of 16 km/h and 20 km/h, respectively. In future investigations of the impact tolerance of lower extremities, other biological parameters should be used for description of the quality of specimens and the structural and mechanical properties of bone. The bone mineral content is one of the parameters that is useful for this purpose.

The compression fracture of the lateral condyle is another damage mechanism that occur at the beginning of the lateral bending in the first damage time-window. This fracture is a result of bone compression in the contact area of femoral and tibial condyles when the preloaded knee joint is forced to bend laterally. This phenomenon was observed only in one test and was related to the low value of the knee bending moment, only 59 Nm. From one test it was impossible to discuss the tolerance level of knee joint condyles to compression which is the result of lateral bending of the lower extremity.

When the initial damage occurs and the knee joint and the lower extremity is forced to bend even after the first damage time-window, other components of the knee joint such as the ACL, the PCL and the joint capsule take over the load absorption and can also be damaged. No damages of the PCL were observed in the present study. The ACL was damaged in three cases and the joint capsule in two cases. The knee bending moment value corresponding to these damages — the ultimate strength of the knee joint to bending moment — is higher than for first sign of damage and is strongly dependent on the configuration of the knee joint. In different configurations, various knee ligaments are activated different ways. In most cases the knee joint resisted larger loads than before the first damage without being further damaged.

The knee lateral bending angle at the time of first sign of damage was 9° (±2) and 11° (±3) for an impact velocity of 16 km/h and 20 km/h, respectively (Table 3 and 4). The results indicate that the knee lateral bending angle at the time of first sign of damage is higher when the lower extremity is exposed to a more rapid load. The difference between the two data sets from tests performed at two velocities used in the present study were not statistically significant on the 5% level. The regression analysis of the 1st PV of KBM and the knee lateral bending angle of two data sets from tests performed at two velocities indicated that the relation between these parameters was not clear. It seems to be difficult to predict the knee bending moment as a function ofknee lateral bending angle.

To describe the consequences of a lateral impact to the lower extremity and the tolerance level of the entire knee joint to the bending moment, the impact force — here the ankle impact force — is not adequate. This force is related to acceleration and the mass of the specimen. Due to test conditions, primarily the leg was accelerated. It was confirmed that the dynamic response of the leg during impact was affected by its mass and was in relation to the mass of the lower extremity. The ankle impact force level — in the impact configuration used — can realistically describe only the ultimate strength of the medial malleolus or distal tibia for direct loads. In all the tests, measured values of the ankle impact force were lower to these, which correspond to the ultimate strength, because no contact damages at the impact point were found. The measured peak ankle impact force was in the range 1.74 (±0.18) kN and 2.41 (±0.51) kN for impact velocities of 16 km/h and 20 km/h, respectively.

A good indicator of the condition of the knee ligaments, foremost the MCL and ACL, was measurements of the valgus-varus and knee laxity. Investigation of the knee laxity before each test gave important information about the condition of the two knee joints of the same cadaver. This kind of testing is necessary to ascertain effects from old injuries to the knee joint and to enable a correct selection of specimen before the test. The knee laxity increment and the valgus-varus increment (Table 2) show good correspondence with the findings in the medical investigations. In pedestrian impact simulation with cadavers these measurements can be very usefully for detecting of ligament damages at the knee joint.

The longitudinal leg rotation, which was observed in the present study, is essential for the development of tensional forces in ligaments in the knee joint. This rotation was observed in most of the tests, in four tests even in the first damage time-window. In these tests no damages to the knee joint were observed. It seems to be that the longitudinal leg rotation is an effect of non-symmetry of the knee joint and the leg-foot segment. During the acceleration of the leg, reaction forces of the knee, the leg and the foot did not act on the longitudinal axis of the lower extremity. The result of these forces was a tendency of the leg to rotate about the longitudinal axis. This phenomenon occurred simultaneously with the impact force acting from the side. The composite result of the longitudinal leg rotation and the impact force was the flexion bending in the knee joint in the sagittal plane. The physiological range of motion for knee flexion is more than 90°. The lateral bending of the knee joint was replaced by a knee flexion, which is a natural movement in the knee joint. This complex phenomenon would protect the knee joint from damages.

The longitudinal leg rotation has not yet been investigated in the published literature, even though many studies on pedestrian side impacts have been reported. Only in one study (Yang and Kajzer, 1992), which presents results from computer simulations of impact to a 3-D model of the knee joint, has this phenomenon been observed. Influence of the sagittal flexion on the reduction of knee joint damages should be clarified in future research.

The limitation of the present study was that the first sign of damage to the entire knee joint was assessed as a rupture of ligaments. In future it is necessary to take into consideration that the human body is not only a mechanical structure in which mechanical failure is the only important sign of injury, but also a living organism capable of many other functions. In many cases even permanent physiological disturbances occur at the cellular level before any rupture can be detected in the tissues. Mechanical failure in tissues is therefore not the only sign — and in the nerve tissue probably not even the most important one — of severe injury. We should therefore focus more on what takes place at the cellular level and not only judge the damage by the disruption of the tissues.

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