



**HAL**  
open science

## Virtual predictive testing of biomechanical effects of impacts on the human leg

Dominique Cesari, François Bermond, Robert Bouquet, Michelle Ramet

► **To cite this version:**

Dominique Cesari, François Bermond, Robert Bouquet, Michelle Ramet. Virtual predictive testing of biomechanical effects of impacts on the human leg. PAM&apos;94 Fourth international conference on virtual predictive testing of industrial prototypes, Oct 1994, PARIS, France. 10 p. hal-02452839

**HAL Id: hal-02452839**

**<https://hal.science/hal-02452839v1>**

Submitted on 23 Jan 2020

**HAL** is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L'archive ouverte pluridisciplinaire **HAL**, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d'enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.

## **Virtual Predictive Testing of Biomechanical Effects of Impacts on the Human Leg**

**Dominique CESARI, François BERMOND, Robert BOUQUET, Michelle RAMET**

INRETS, Institut National de Recherche sur les Transports et leur Sécurité

109 Avenue Salvador Allende, case 24, 69675 BRON cédex, FRANCE

### **ABSTRACT**

In car/pedestrian accident mathematical simulations, it is desirable to extend existing rigid body human model towards deformable finite element models. Thereby a wider range of front car structure/pedestrian interactions can be covered. Even though some injuries to pedestrian are due to contact with the ground, research concerning the severity of injuries due to the car contact, related to the bumper and the bonnet height and stiffness, and to the speed of the vehicle, are being carried out.

In a previous study a finite element model of the pedestrian knee-joint in lateral impact was expanded. This paper presents a modelling of a human leg in lateral impact for use in an explicit finite element code. The articulation between each condyle of the femur and the corresponding tuberosity of the tibia is mainly described. The bones entering into the formation of the knee-joint are the condyles of the femur above, the head of the tibia below. The bones are connected together by ligaments, some of which are placed on the exterior (e. g. ; lateral and medial collateral...) of the joint, while others occupy its interior (e. g. ; anterior and posterior crucial...).

This model will be impacted to estimate the severity of leg lesions and to predict the risk of leg injuries in car/pedestrian accidents. The effect of the impactor stiffness, velocity and height will be evaluated.

Experimental results from static tests and impact tests, to characterise the mechanical behaviour of each part, like the ligaments or the bones, and the whole kinematics of the knee, serve as a basis for the validation of the model.

### **FOREWORD**

Passive safety improvement has been a first priority as well for the car manufactures as for governments in developed countries. This has generated a large amount of research and developments.

Up to now the protection offered by cars and by safety devices is evaluated in tests using an anthropomorphic instrumented dummy.

The development of numerical simulation has made possible the use of "mathematical dummies". Nevertheless a dummy whatever it is, mechanical or mathematical has limited representation of human living. For those reasons, the protection developed using a dummy cannot be optimized for human living. There is then, a need to develop a realistic mathematical model of human body enable to predict the effect of impacts in term of injury risk. Compared to dummy model a finite element model of the human body allow to evaluate the effectiveness of protection devices for different occupants.

Biomechanical researches have shown that human tolerance varies according to the anthropometry, the age, and he sex. By changing values of parameters related to the above characteristics it is then possible to explore widely the level of protection. The development of a finite element model of the human body is an ambitious project, and in the field of crash protection it would give usable results if it is coupled with a model of deformable car.

However without waiting for this complete model it is possible to use a model of a human body part in specific and limited application. The general aim of human leg finite element model is to able to analyze the effect of a car front impact on a human leg, in complement to a standart test using a mechanical leg.

### **INTRODUCTION**

In this paper, statistical analyses for pedestrians accidents, the description of the car/pedestrian accidents, an overview of the functional anatomy of the lower limb, the main car/pedestrian leg injuries and the majority crash simulation are showed. Then the methodology for the modelling, the validation of the human leg model and results are presented. The code used for the mathematical simulations is the three dimensional non-linear explicit Finite Element code PAM-CRASH developed by ESI (Engineering Systems International S.A., France).

#### **Description of the car/pedestrian accident**

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.

Car/pedestrian accident could be described firstly by the impacts against the front of the car, bumper and the bonnet edge for the lower body region, mainly for the leg ; secondly the impacts against the bonnet or the windscreen for the head. Finally there is the contact with the ground for the whole body.

The shape of the front of the car has an influence on the pedestrian kinematics, not only on its trajectory but also on the velocity of each body segment. The kinematics of a pedestrian is a major factor in determining the location, the occurrence

of contacts between the different body segments against a vehicle, the ground (Grösch, 1989), and also the impact velocity of these contacts (EEVC, 1982).

### **Car/pedestrian accident assumptions**

Then some years ago INRETS decided to go further and try to understand the question of pedestrian and leg injuries. A program has been started with different cars, impacting cadavers and dummies. This program has proceeded step by step. The first step was a full scale cadaver test in which the deformation of the leg at the time of the impact was observed from high speed films. What was interesting was that when this deformation of the leg occurred around the front of the car, the upper part of the body, which was free and not suspended at that time, did not move. It moved later. So the leg injuries can be considered quite separately from those to other body parts.

Also, due to the complexity of car/pedestrian accident description, in this paper we focus only on the first part of the car/pedestrian accident which is the contact between the front of the car and the pedestrian leg.

### **Functional anatomy of the leg (Gray, 1977)**

**Leg Bones** -The skeleton of the leg (lower limb) consists mainly of two parts : the upper leg (thigh) and the lower leg. The thigh is that portion of the lower extremity which is situated between the pelvis and the knee.

The femur is the longest, and strongest bone in the skeleton, and almost perfectly cylindrical in the greater part of its extent. The femur is divisible into a shaft and two extremities. The shaft of the femur is a cylinder of compact tissue, hollowed by a large medullary canal. The cylinder is of great thickness and density in the middle third of the shaft, where the bone is narrowest and the medullary canal well formed ; but above and below this the cylinder gradually becomes thinner, owing to a separation of the layers of the bone into cancelli, which project into the medullary canal and finally obliterate it, so that the upper and lower ends of the shaft, and the articular extremities more especially, consist of cancelled tissue invested by a thin, compact layer.

The skeleton of the lower leg consists of three bones : the patella, the tibia and the fibula.

The patella is a flat triangular bone, situated at the anterior part of the knee-joint. The tibia is situated at the front and inner side of the leg, and, excepting the femur, is the longest and largest bone in the skeleton. It is prismoid in form, expanded above, where it enters into the knee-joint, more slightly enlarged below. It presents a shaft and two extremities and its structure is like that of the femur. The fibula is situated at the outer side of the leg. It is the smaller of the two bones of the lower leg, and, in proportion to its length, the most slender of all the long bones : it is placed on the outer side of the tibia, with which it is connected above and below. Its upper extremity is small, placed toward the back of the head of the tibia and below the level of the knee-joint, and excluded from its formation : the lower extremity inclines a little forward, so as to be on a plane anterior to that of the upper end, projects below the tibia, and forms the outer ankle. It also presents a shaft and two extremities and its structure is like of the femur.

**Knee joint** - The knee-joint must be regarded as consisting of three articulations in one : one between each condyle of the femur and the corresponding tuberosity of the tibia, and one between the patella and the femur. The bones entering into the formation of the knee-joint are the condyles of the femur above, the head of the tibia below. The bones are connected together by ligaments, some which are placed on the exterior of the joint, while others occupy its interior.

The main ligaments in the knee joint are : the External Ligaments are the Lateral Collateral Ligament and the Medial Collateral Ligament ; which are stressed when the leg is stretched and relaxed when the knee is flexed ; the Interior ligaments are the Anterior Crucial Ligament and the Posterior Crucial Ligament ; which are always stressed.

**Muscles** - The main muscles around the knee-joint are : in front and at the sides, the Quadriceps extensor ; on the outer side, the tendons of the Biceps and the Popliteus ; on the inner side, the Sartorius, Gracilis, Semitendinosus and Semimembranosus ; behind, an expansion from the tendon of the Semimembranosus.

**Meniscus** - The meniscus are two crescentic lamellae which serve to deepen the surface of the head of the tibia, for articulation with the condyles of the femur. The circumference of each cartilage is thick, convex, and attached to inside the capsule of the knee.

### **Car/pedestrian lower limb injuries**

In car/pedestrian accident, injuries are caused by the impacts against the front of the car, bumper and the bonnet edge for the lower body region, mainly for the leg ; and the impacts against the bonnet or the windscreen for the head. Others injuries are also due to the contact with the ground.

The severity of injuries due to the car contact are related to the bumper and the bonnet height and stiffness, and the speed of the car. All these injuries were not fatal, but they led to very severe disabilities and impairments generally for a long

duration. In some cases, they demonstrated an irreversible character, especially when the speed of the involved vehicle was higher than 30 km/h. The main injuries observed in adults (Manoli, 1986) relate to bone segment fractures, femur-tibia or fibula, articular troubles especially at the knee ligament level and soft tissues of the whole leg.

Oftentimes the tibia and/or fibula are the site of initial contact between the automobile and the pedestrian. The effect of bumper heights on the location of the fracture has been well summarized (Ashton et al, 1983). The injuries to the bones themselves may be severe. The skin and muscle tissue surrounding them are frequently damaged severely as well. Knee injuries consist of either injuries to the knee ligaments and soft tissues, or to the bones contributing to the articulation, or to both.

Knee ligament are the results of a bending force applied to the joint. In automobile-pedestrian interactions, valgus stress may tear the medial structure. A valgus stress injury may result in injury to the medial collateral ligament of the knee and medial knee cartilage. These ligament tears may be incomplete or complete, with total disruption of the fibres. Also commonly associated is the injury to the anterior crucial ligament and, on rare occasions, the posterior crucial ligament as well. If all of these structures and the capsule of the joint are also torn, a knee dislocation may result.

Severe injuries (complete tears) of the ligaments are best treated surgically with repair of the ligaments by direct suture after the initial evaluation is completed. The anterior crucial ligament is frequently difficult to repair as it is often torn in the mid-substance of its fibres.

Fractures of the joint may also occur with or without ligament injury. Valgus stress injury may result in fracture to the lateral tibial condyle. In general, if there is displacement, these are operated on and open reduction and internal fixation is performed.

It should be stressed that the knee ligament and bone articulation fractures are extremely severe injuries. Furthermore, because of their long term consequences, even if they are not very frequent, ligament injuries must be considered very carefully.

### **Majority crash simulation**

There are many ways of pedestrian accident simulation ; the full-scale tests in which a vehicle hits a dummy or a cadaver, the rig tests in which an impactor simulating a segment of human body hits a chosen area of a vehicle or an impactor simulating a chosen area of a vehicle hits a segment of human body, subsystem test methods to evaluate Pedestrian protection, and, mathematical modelling of a vehicle hitting a pedestrian.

**Mechanical models** - For the evaluation of car-front aggressiveness to pedestrians in a car/pedestrian collision, subsystems and full pedestrian dummies have been considered for a leg-to-bumper impact test.

The Rotationally Symmetrical Pedestrian Dummy (RSPD), developed at Department of Injury Prevention, Chalmers University of Technology in Sweden and INRETS in France, is a pedestrian dummy equipped with a system for measuring the moments and forces in the lower extremities, especially at the knee joint (Aldman et al, 1985).

A mechanical representation of the lower limb in form of an impactor with deformable knee joint developed by INRETS is representative of a subsystem (Cesari et al, 1991).

Four mechanical substitutes of a pedestrian were used to test the influence of different car-front shapes and dummy parameters on the results ; the leg of the Rotationally Symmetrical Pedestrian Dummy (RSPD), a Hybrid-II pedestrian dummy, a modified Hybrid-II pedestrian dummy equipped with a steel bar serving as knee joint, and a RSPD - Hybrid-IIP combined dummy in which the lower part of the RSPD and the upper part of the Hybrid-IIP were connected by a joint in such a way that the movements of the upper part were similar to those in cadaver tests (Ishikawa et al, 1992).

**Mathematical models** - A 2D mathematical simulations of the pedestrian leg in lateral impact were conducted by INRETS (Bermond et al, 1992) with the rigid body program Madymo (TNO). Results of this model were compared with those obtained with the instrumented mechanical leg, developed by INRETS.

A 3D pedestrian knee joint model was developed as a first step in a new description of the whole pedestrian body for computer simulations (Yang et al, 1992) with Madymo. The new developed model was verified with results from tests with biological material previously performed at the Department of Injury Prevention, Chalmers University of Technology in Sweden.

A mathematical breakable leg model was developed and implemented into the pedestrian lower extremity model (Yang et al, 1993). The leg model, described with Madymo (TNO) consists of two rigid-body elements connected by a fracturable joint. The results showed that the modified model gave a higher biofidelity than did the previous model with the undeformable representation of the leg segments.

A pedestrian model was created to be used with the Crash Victim Simulation (CVS). The model consists of fifteen segments connected by fourteen joints. The geometry and the characteristics of the body segments, and the mechanical properties of

the joints are based on available anthropometrical and biomechanical data. Results were compared with cadaver tests (Ishikawa et al, 1993).

A 2D dynamic anatomical model of the human knee joint (Rahman et al, 1993) simulate its response under impact. The knee joint is modeled as two rigid bodies, representing a fixed femur and a moving tibia, connected by 10 non-linear springs. Two springs represented the anterior and the posterior fibers of each of the anterior and the posterior crucial ligament, four springs represented the medial collateral ligament, and one spring represented each of the lateral collateral ligament and the posterior part of the capsule. Knee response was determined under sudden rectangular pulsing posterior forces applied to the tibia and having different amplitudes and durations. It is hard to compare this model calculation with other data because of the limited amount of experimental data available in the literature.

A three body segment dynamic model of the human knee (Tümer et al, 1993) includes tibio-femoral and patello-femoral articulations, and anterior crucial, posterior crucial, medial collateral, lateral collateral, and patellar ligaments. A specially developed human knee animation program is utilized in order to fine tune some model parameters. Numerical results are presented for knee extension under the impulsive action of the quadriceps femoris muscle group to simulate a vigorous lower limb activity such as kicking. The results are discussed and compared with limited data reported in the literature.

These car/pedestrian accident mathematical simulations were carried out with the rigid body program Madymo or CVS. A significant shortcoming of such an approach, however, is the need to provide experimental force deflection data as input for the contact models. In addition the geometry is met only poor, e.g. by ellipsoids. Finite Element models, on the other hand, can potentially cover a much wider range of loading situations. As the level of modelling is fundamentally different, such models are based essentially on material properties and the geometry of the surrogate to be discretized.

A finite element model of the pedestrian knee-joint in lateral impact (Bermond et al, 1993) was compared with cadaver tests (Kajzer, 1990).

In this paper this study was expanded to the lower limb.

## METHODOLOGY

### Lower limb model assumptions

To analyse the complexity of the problem several assumptions are made to simulate the behaviour of the lower limb in lateral impact :

- only the two bones are taken into account, the femur and the tibia. Only the cortical bone is described, because the spongy tissue is less stiff. The patella and the fibula do not influence significantly the response.
- only the joint of the knee is described with the posterior and anterior crucial ligament, and the medial and lateral collateral ligament.

The others ligaments and tendons and muscles are not described. At the beginning of the impact however the muscles are stressed, we suppose they do not change the response a lot, because they are less stiff than the four ligaments we have chosen.

The meniscus are not yet represented, because mechanical properties data are not easily available.

### Geometry of the lower limb model

The bones are described in 3D space. The shape of the shaft the femur and the tibia is reconstructed from human leg X-ray scanner. The geometry is discretized with shell elements. The mesh of the two extremities of the femur and the tibia are obtained with an ultrasonic device (GP8 3D from Science Accessories Corporation). A mesh is drawn by hand on real bones and nodes are reached with the stylet which gives on a file the three dimensional coordinates of the node. The femur mesh is discretized with around 600 nodes and 650 elements, and the tibia mesh is discretized with 500 nodes and 550 elements.

The ligaments are described as a 1D linkage element with non-linear behaviour.

The figure 1 shows the human lower limb model in anterior view, side view and posterior view.

### Material properties of the lower limb model components

The material properties of the lower limb model components are found in the literature (Viano, 1986 and Skalak, 1987).

**Mechanical properties of bones** - The characteristic of the bone (Burstein et al, 1976, Keaveny et al, 1993) (table 1) is viscoelastic quality. Cortical bone is similar in properties to other fibrous materials such as wood, and has substantially more compliance than engineering materials such as metals, but more rigidity than spongy bone. Bone is one of the most rigid biological materials in the body and has a significantly greater stress carrying capacity than soft tissues which are frequently used to link long bones through joints or cover the musculoskeletal system.

Cortical bone	Femur	Tibia
Density : kg/m <sup>3</sup>	1900	1900
Elastic modulus : N/m <sup>2</sup>	17.6 10 <sup>9</sup>	18.4 10 <sup>9</sup>

Plastic modulus : N/m <sup>2</sup>	0.754 10 <sup>9</sup>	1.2 10 <sup>9</sup>
Yield stress : N/m <sup>2</sup>	0.12 10 <sup>9</sup>	0.13 10 <sup>9</sup>
Ultimate stress : N/m <sup>2</sup>	0.14 10 <sup>9</sup>	0.15 10 <sup>9</sup>
Poisson's ratio	0.326	0.326

*Table 1 : Mechanical properties of bones.*

**Mechanical properties of the ligament** - The main function of ligament (Herzberg et al, 1981, Race et al, 1993, and Woo et al, 1993) (table 2) is to provide stability to joints and limit their range of motion and is to resist tensile loading either due to muscular contraction or loads tending to displace the joint.

Mechanical properties of ligament have mainly been determined by tensile testing of isolated tissues.

Ligament	Lateral collateral	Medial collateral	Anterior Crucial	Posterior Crucial
Elastic modulus : N/m <sup>2</sup>	15 10 <sup>6</sup>	15 10 <sup>6</sup>	30 10 <sup>6</sup>	35 10 <sup>6</sup>
Ultimate stress : N/m <sup>2</sup>	400 10 <sup>6</sup>	200 10 <sup>6</sup>	200 10 <sup>6</sup>	100 10 <sup>6</sup>
Rupture force : N	3000	3000	6000	6000
Ultimate Strain at rupture : %	30	40	60	60
Density : kg/m <sup>3</sup>	1000	1000	1000	1000
Poisson's ratio	0.3	0.3	0.3	0.3

*Table 2 : Mechanical properties of the crucial and collateral ligaments.*

### **Impactor assumptions**

The form of the impactor is a half cylinder. It represents an approach of the bumper shape. The geometry is discretized with shell elements. The material properties of the iron are chosen. The impactor weights 16 kg.

### **MODEL VALIDATION**

Experimental results from static tests and impact tests, to characterise, the mechanical behaviour of each part, like the ligaments or the bones, and the whole kinematics of the lower limb, will serve as a basis for the validation of the model. Two INRETS laboratories (Laboratory of Impacts and Biomechanics (L.C.B.) at Bron and the Applied Biomechanics Laboratory (L.B.A.) of Marseille) and a laboratory at Chalmers University (Department of Injury Prevention (D.I.P.) at Göteborg) are linked by a programme. This project was to analyse the effects of shearing loads (Kajzer, 1990) and bending moment (Kajzer, 1993) applied laterally to the human knee joint.

This experimental study was carried out at L.B.A. (INRETS University of Aix/Marseille-II Associated Research Unit) at the North Faculty of Medicine of Marseille. 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.

### **Effect of shearing loads**

Nineteen tests with lower limbs from cadavers were carried out : nine for an impact speed of 15 km/h and 10 at 20 km/h. Dynamic testing of the leg was made using an impactor which was propelled by sandows. A specially designed impact arm for doing shearing load, equipped with force transducers was mounted in front of the impactor. Figure 2 shows the initial of the tests procedure.

### **Effect of bending moment**

Seventeen test were carried out under dynamic conditions : seven at a velocity of 16 km/h and ten at 20 km/h. A specially designed impact face for doing bending moment was used. Initial test conditions are showed in figure 3.

### **Tests**

To perform analysis of every test, two high speed cameras (1000 frames/s) were used during the tests and six targets were fixed on femur and tibia to permit high speed films to give the information about the kinematics of the leg. The dynamic response was measured with several force transducers.

### **Validation**

To validated our model we compare the results from the tests with those from our model without forgotten our lower limb joint model assumptions.

## RESULTS FROM THE LEG MODEL

The results from the computer simulations are shown in figure 4 to 8, with the units from the System International.

The upper part and the external side of the tibia is impacted at 2.33 m/s.

On the figure 4 the contact force magnitude between the impactor and the tibia is maximum at around 0.005 s.

The figure 5 shows the internal energy (due to the strain of the material), the kinetic energy (due to the displacement of the material), of the tibia, the femur and the ligaments. The ligaments accumulate more internal energy than kinetic energy, and the opposite appear for the bones. The tibia accumulate more kinetic energy and internal energy than the femur.

The axial force in the ligaments are shown on figure 6 and are filtered using SAE channel. During the first 0.01 s, the medial collateral ligament are more stressed in tension than the anterior crucial ligament, and the lateral collateral ligament are more stressed in compression than the posterior crucial ligament. After, from 0.01 s until 0.016 s the posterior and the lateral collateral ligament is stressed in tension, and the anterior and the medial collateral ligament is stressed in compression.

The initial position of the human lower limb model with the impactor model is presented at figure 7. On the same figure the deformed shapes of the leg model are showed at 0.005 s, 0.0055 s, 0.006 s and 0.0065 s.

For each direction x, y and z, the displacement, the velocity and the acceleration curves for each node of the model are calculated and also the magnitude. Theses curves are filtered using SAE channel class from 1000 Hz to 60 Hz. There are peak values corresponding at the impact. The maximum effective surface stress and the Von Mises stress are also calculated for each element. The maximum effective surface stress is concentrated around the area of the impact at 0.004 s and after near the ligament insertion at 0.008 s.

On figure 8 the velocity vectors are plotted at 0.005 s, 0.0055 s and 0.006 s. The velocity vector field, at 0.005 s, is quite well regular distributed for all the lower leg. Then, at 0.0055 s, it is concentrated on the bottom of the tibia and at the knee. After at 0.006 s, it is quite regular distributed for all the lower limb.

## CONCLUSIONS

The aim of this paper was to create a second step for a human lower limb model in lateral impact. The first step in a previous study concerned the human knee joint model. This approach with Finite Element Method gives information about the internal mechanical behaviour, the kinematics, the displacement, the velocity, the acceleration, the maximum effective surface stress, the Von Mises stress and the contact force between the impactor and the tibia.

The model is not very well validated, and we will reduce several assumptions such as the response of the soft tissue and the muscle, or the whole shape of the leg. The mesh will be improved.

We still have to simulate other use conditions of the model as a function of the stiffness, the height the speed or the energy of the impactor. Both aspects of modelling and experimental approach still require a lot of tests.

The leg model developed for the specific problem of pedestrian leg injury prediction may contribute to the development of full model of the human body.

## ACKNOWLEDGEMENTS

The authors would like to express their appreciation to Yves CAIRE, Gérard GOUTELLE, Pierre LAPELERIE, Michel MARAIS for doing such a fine job in collecting data, photos, measurements, in these experiments. This research could not have been done without their work. The Functional anatomy of the knee was done by M. BOUALLEGUE, A. MORIN and J.P.H. NEIDHARDT from the "Laboratoire d'Anatomie de la Faculté de Médecine de Lyon Nord".

## REFERENCES

**B. Aldman, J. Kajzer, D. Cesari, R. Bouquet, R. Zac**, *A New Dummy for Pedestrian Test*, Proceedings of the tenth International Technical Conference on Experimental Safety Vehicles, Oxford, England, July 1-4, 1985, pp 176-185.

**S.J. Ashton and G.M. Mackay**, *Benefits from Changes in Vehicle Exterior Design Field Accident and Experimental Work in Europe. In pedestrian impact injury & assessment*, Warrendale, PA, Society of Automotive Engineers, pp 205-223, 1983.

**F. Bermond, D. Cesari, F. Alonzo, M. Matyjewski**, *Mathematical Simulation of the Pedestrian Leg in Lateral Impact*, Proceedings of the 1992 International IRCOBI Conference on the Biomechanics of Impacts, Verona, Italy, September 9-11, 1992, pp 61-72.

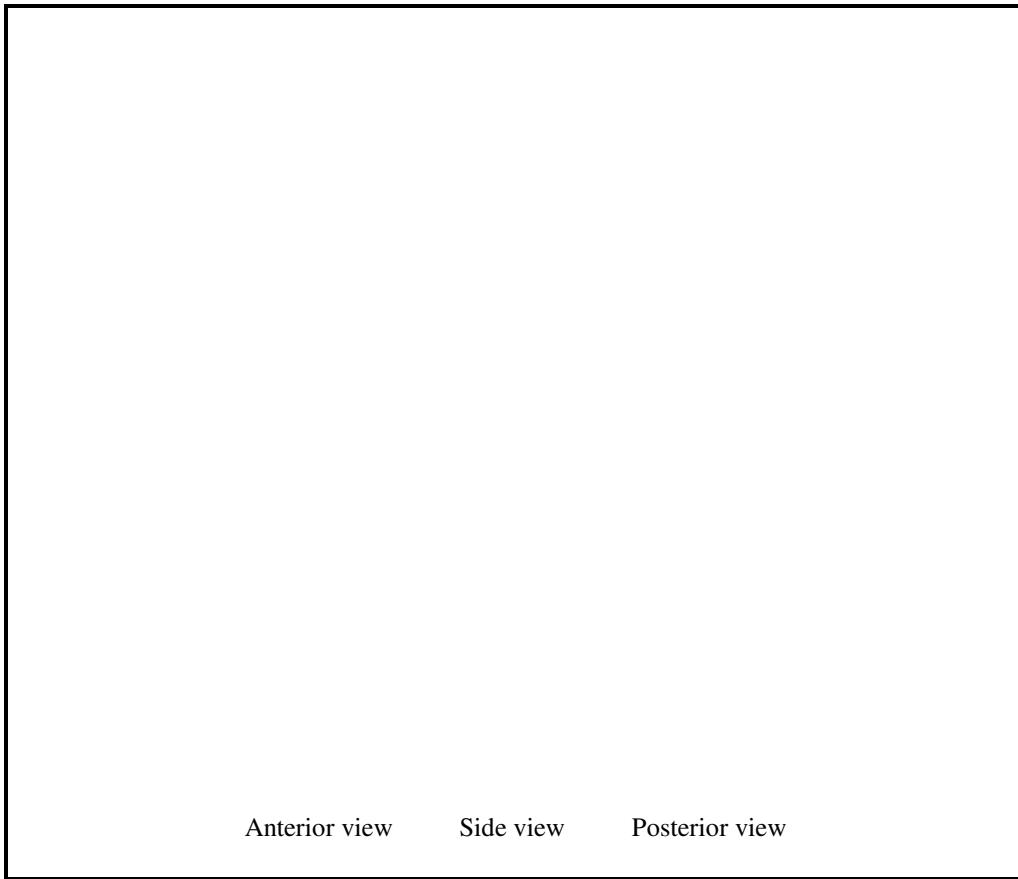
**F. Bermond, M. Ramet, R. Bouquet, D. Cesari**, *A Finite Element Model of the Pedestrian Knee-joint in Lateral Impact*, Proceedings of the 1993 International IRCOBI Conference on the Biomechanics of Impacts, Eindhoven, The Netherlands, September 8-10, 1993, pp 117-129.

**H. Burstein, T. Reilly, M. Martens**, *Ageing of Bone Tissue : Mechanical Properties*, Journal of Bone and Joint Surgery, Vol. 58-A, No. 1, pp 82-86, January 1976.

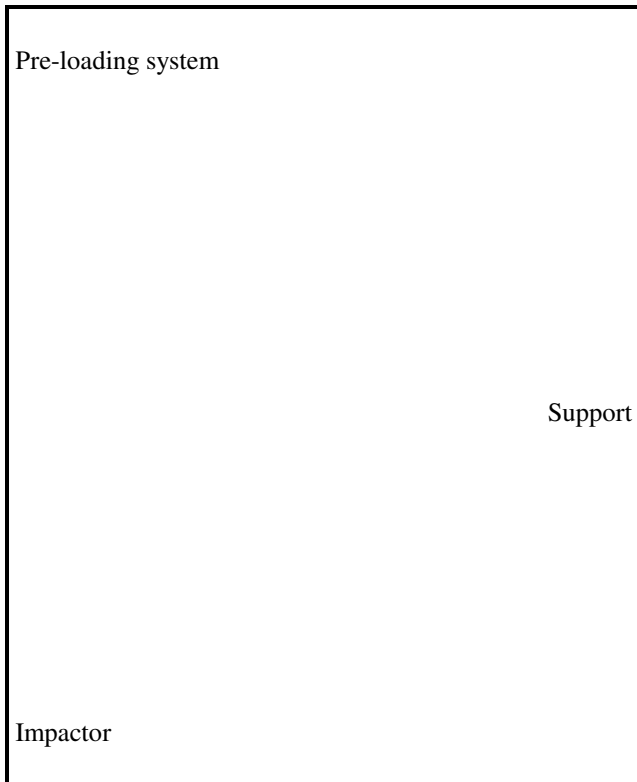
**J. Castaing, Ph. Burdin**, *Anatomie Fonctionnelle de l'Appareil Locomoteur : le Genou*, Edit. E.P.R.I., Paris, 1980.

- D. Cesari, F. Alonzo, M. Matyjewski**, *Subsystem Test for Pedestrian Lower Leg and Knee Protection*, Proceedings of the thirteenth International Technical Conference on Experimental Safety Vehicles, Paris, France, November 4-7, 1991, 91-S3-O-08, pp 310-317.
- Y.C. Fung**, *Biomechanics : Mechanical Properties of Living Tissues*, Ed. Springer Verlag, 1981.
- H. Gray**, *Anatomy, Descriptive and Surgical*, Ed. Bounty Books, 1977.
- L. Grösch, W. Heiss**, *Bumper Configurations for Conflicting Requirements : Existing Performance Versus Pedestrian Protection*, Proceedings of the Twelfth International Technical Conference on Experimental Safety Vehicles, Göteborg, Sweden, May 29- June 1, 1989, pp 1266-1273.
- G. Herzberg, P. Fischer, G. Gonon, J.P. Carret, M. Journot, N. Uezaki**, *De la Résistance à la Traction des Ligaments et Tendons Péri-articulaires du Genou*, Bulletin de l'association des anatomistes, Vol. 188, No. 65, pp 113-118, 1981.
- H. Ishikawa, J. Kajzer, K. Ono, M. Sakurai**, *Simulation of Car Impact to Pedestrian Lower Extremity*, Proceedings of the 1992 International IRCOBI Conference on the Biomechanics of Impacts, Verona, Italy, September 9-11, 1992, pp 1-12.
- H. Ishikawa, J. Kajzer, G. Schroeder**, *Computer Simulation of Impact Response of the Human Body in Car Pedestrian Accidents*, Proceedings of the thirty-seventh Stapp Car Crash Conference, San Antonio, Texas, November 1-3, 1993, pp 235-248.
- J. Kajzer, C. Cavallero, S. Ghanouchi, J. Bonnoit, A. Ghorbel**, *Response of the Knee Joint in Lateral Impact : Effect of Shearing Loads*, Proceedings of the 1990 International IRCOBI Conference on the Biomechanics of Impacts, Bron, France, September 12-14, 1990, pp 293-304.
- J. Kajzer, C. Cavallero, J. Bonnoit, A. Morjane, S. Ghanouchi**, *Response of the Knee Joint in Lateral Impact : Effect of Bending Moment*, Proceedings of the 1993 International IRCOBI Conference on the Biomechanics of Impacts, Eindhoven, The Netherlands, September 8-10, 1993, pp 105-116.
- T.M. Keaveny, W.C. Hayes**, *A 20-Year Perspective Mechanical Properties of Trabecular Bone*, Journal of Biomechanical Engineering, November 1993, Vol. 115, pp 534-542.
- A. Manoli**, *Adult Pedestrian Injuries - Medical Aspects*, Symposium on Biomechanics and Medical Aspects of Lower Limb Injuries, San Diego, California, October 29-30, 1986, SAE P-186, 861928, pp 109-114.
- F.H. Netter**, *The CIBA collection of Medical illustrations*, Vol. 8, part 1, CIBA-GEIGY Corporation, Summit, New Jersey, USA, 1987.
- A. Race, A.A. Amis**, *The Mechanical Properties of the Two Bundles of the Human Posterior Crucial Ligament*, Vol. 27, No. 1, Journal of Biomechanics, 1994, pp 13-24.
- E.A. Rahman, M.S. Hefzy**, *A Two-Dimensional Dynamic Anatomical Model of the Human Knee joint*, Journal of Biomechanical Engineering, November 1993, Vol. 115, pp 357-365.
- R. Skalak, S. Chien**, *Handbook of Bioengineering*, Mc Graw-Hill Book Company, 1987.
- S.T. Tümer, A.E. Engin**, *Three-Body Segment Dynamic Model of the Human Knee*, Vol. 115, Journal of Biomechanical Engineering, November 1993, pp 350-356.
- C. Viano**, *Biomechanics of Bone and Tissue : a Review of Material Properties and Failure Characteristics*, Symposium on Biomechanics and Medical Aspects of Lower Limb Injuries, San Diego, California, October 29-30, 1986, SAE P-186, 861928, pp 33-63.
- S.L-Y. Woo, G.A. Johnson, B.A. Smith**, *Mathematical Modeling of Ligament and tendons*, Journal of Biomechanical Engineering, November 1993, Vol. 115, pp 468-473.
- J. Yang, J. Kajzer**, *Computer Simulation of Impact Response of the Human Knee Joint in Car-Pedestrian Accidents*, Proceedings of the thirty-sixth Stapp Car Crash Conference, Seattle, Washington, November 2-4, 1992, pp 203-217.
- J. Yang, C. Rzymkowski and J. Kajzer**, *Development and Validation of a Mathematical Breakable Leg Model*, Proceedings of the 1993 International IRCOBI Conference on the Biomechanics of Impacts, Eindhoven, The Netherlands, September 8-10, 1993, pp 175-186.

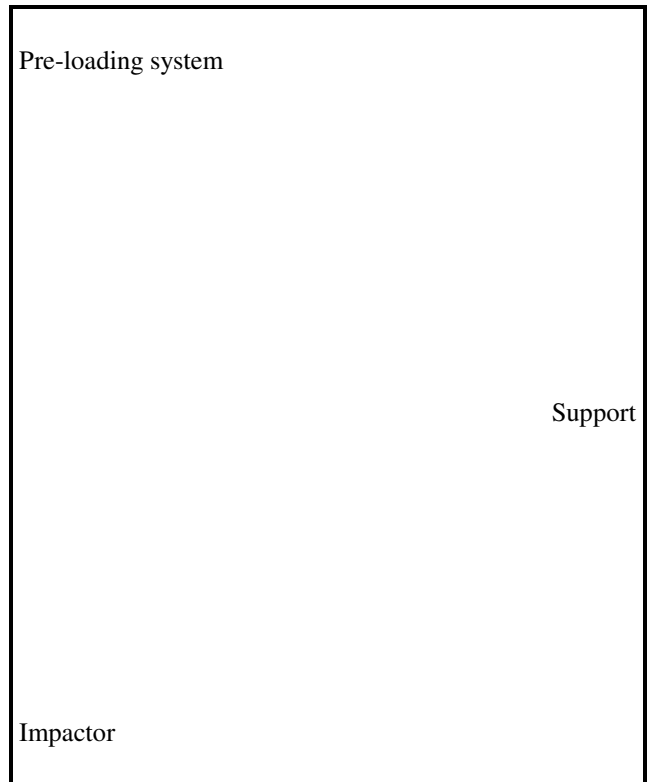




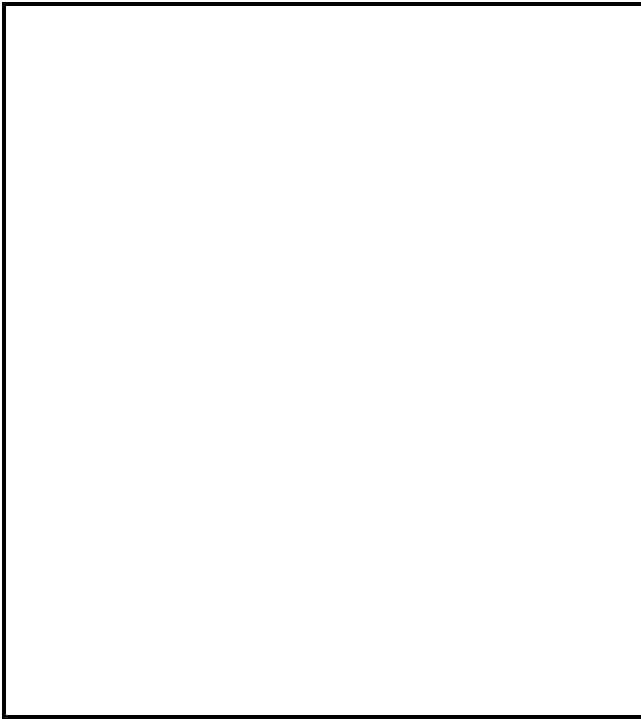
*Figure 1 : Human lower limb model.*



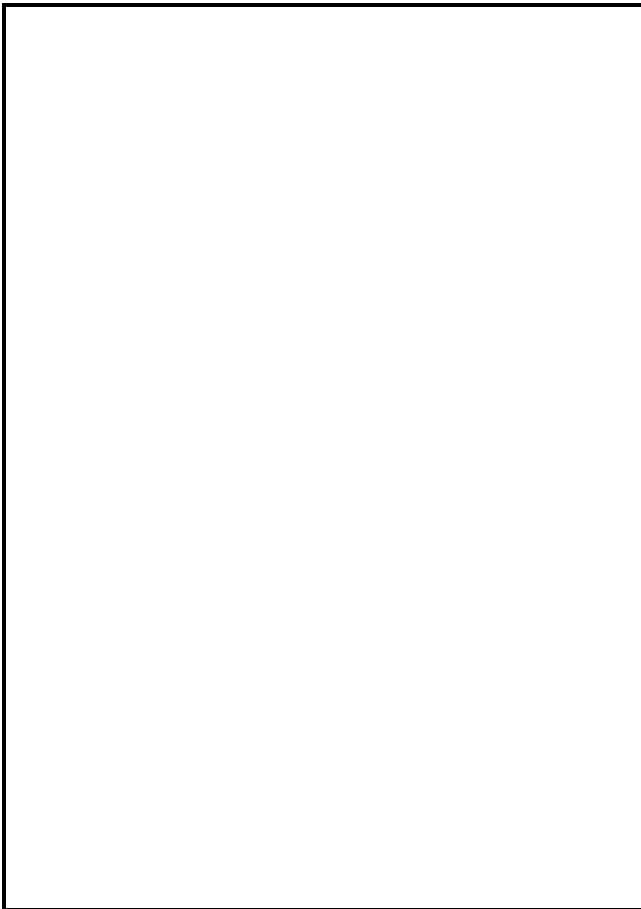
*Figure 2 : Human leg in lateral impact (Kajzer, 1990), Shearing load test.*



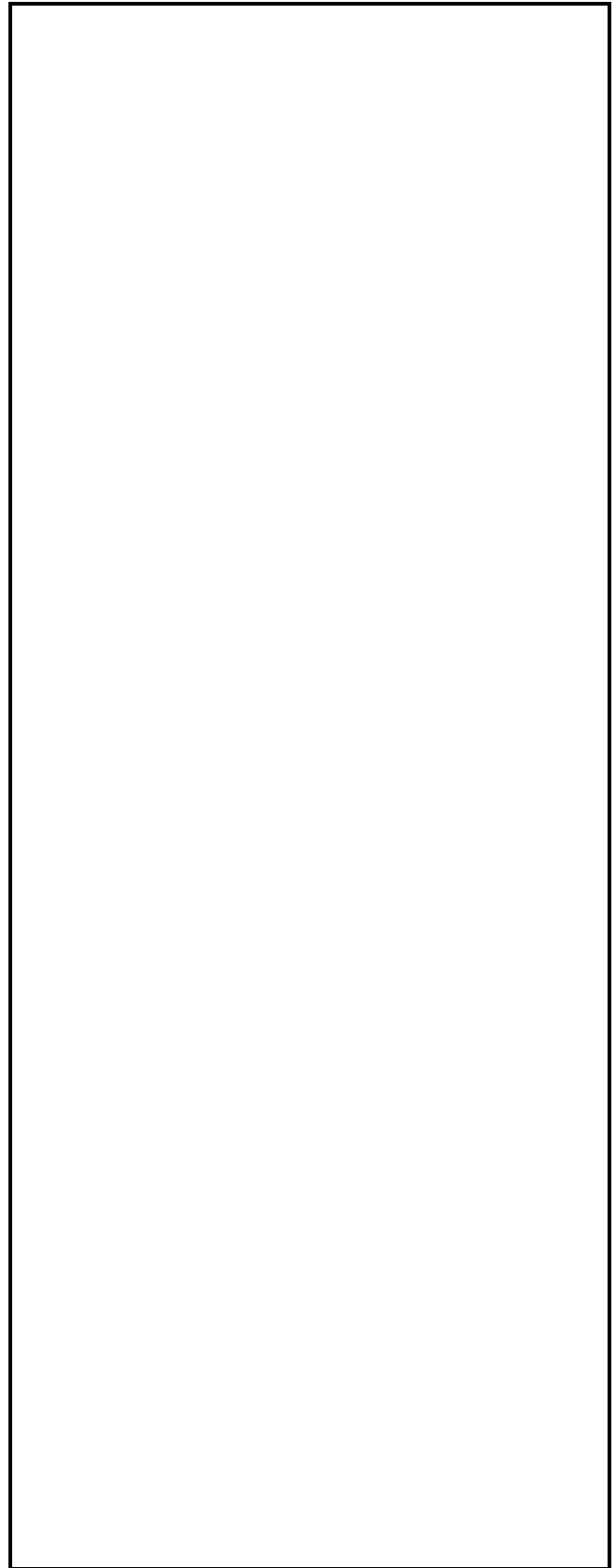
*Figure 3 : Human leg in lateral impact (Kajzer, 1993), Bending moment test.*



*Figure 4 : Contact force, between impactor and tibia.*



*Figure 6 :Axial force in the ligament,  
No. 1 : medial collateral, No. 2 : lateral collateral,  
No. 3 : anterior crucial, No. 4 : posterior crucial.*



*Figure 5 : Energies of material,  
Material No. 2 : femur, material No. 5 : tibia, material  
No. 8 : ligaments.*

T = 0.0000 T = 0.0050 T = 0.0055 T = 0.0060 T = 0.0065

Figure 7 : Lower leg impact test, kinematics.

T = 0.0050 T = 0.0055 T = 0.006

Figure 8 : Velocity vector.