A FINITE ELEMENT MODEL OF THE PEDESTRIAN KNEE-JOINT IN LATERAL IMPACT

F.BERMOND, M.RAMET, R.BOUQUET, D.CESARI

INRETS, 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 pedestrian 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.

This paper presents a modelling of a human knee-joint 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 (for example ; lateral and medial collateral...) of the joint, while others occupy its interior (for example ; anterior and posterior crucial...).

This model will be impacted to estimate the severity of knee joint lesions and to predict the risk of knee 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.

Introduction

Pedestrians are not protected when they are crossing a street and are impacted by vehicles. Every year thousands of pedestrians are killed or injured in road traffic accidents in the world. The number of people injured in Europe, USA and Japan decreased from 33,000 in 1970 to 18,000 in 1986 (Vallée et al, 1989). Since then, the number of pedestrian fatalities in traffic accidents has been almost stable. The number of killed pedestrians compared to the total number of killed road users varied from 13 percent in the Netherlands (EEVC/CEVE, 1982) to 15 percent in the USA, to 18 percent in Europe (Vallée et al, 1989), to 30 percent in Japan (Ishikawa et al, 1991) and 45 percent in Dehli in India (Sarin et al, 1990).

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.

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 (RPSD), a Hybrid-II pedestrian dummy, a modified Hybrid-II pedestrian dummy equipped with a steel bar serving as knee joint, and a RPSD - 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).

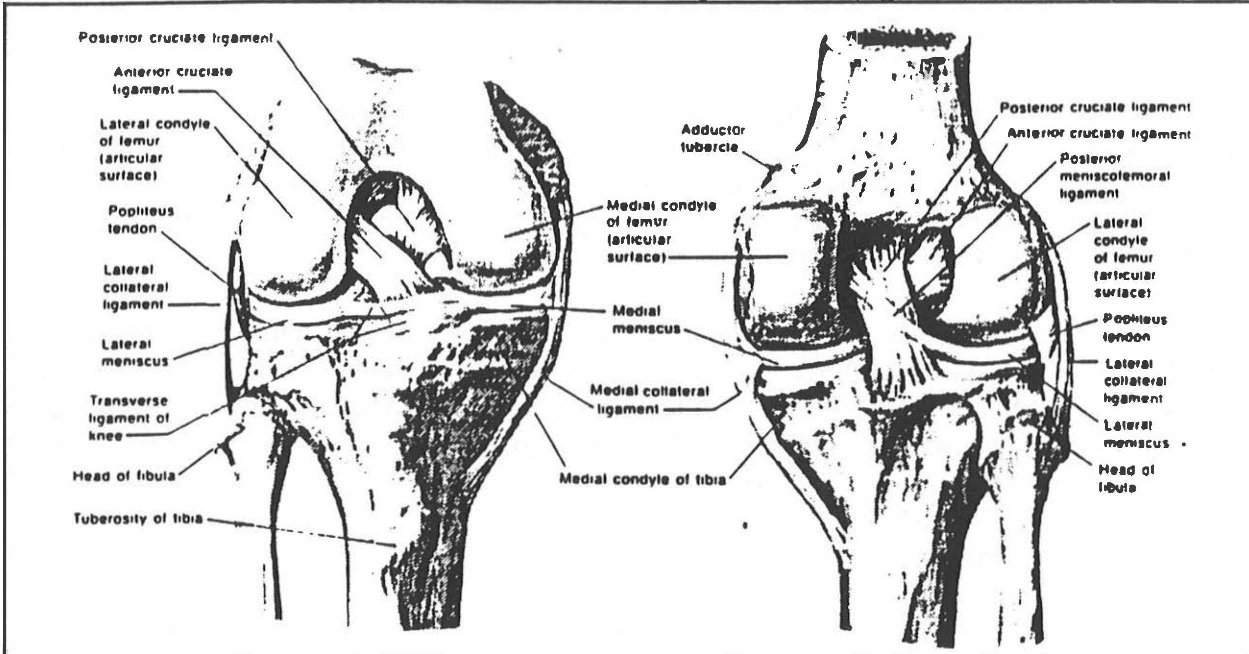
2D mathematical simulations of the pedestrian leg in lateral impact were conducted by INRETS (Bermond et al, 1992). Results of this model were compared with those obtained with the instrumented mechanical leg, developed by INRETS.
A 3D pedestrian knee joint model was developed as a first step in a new description of the whole pedestrian body for computer simulations (Yang et al, 1992). 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.
These car/pedestrian accident mathematical simulations were carried out with the rigid body program Madymo. 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.

In this paper an overview of the anatomy of the knee joint and the main car/pedestrian knee injuries is showed ; and the modelling and validation of the knee of a human is presented. The code used for the simulations is the non-linear explicit Finite Element code PAM-CRASH developed by ESI (Engineering Systems International S.A., France).

Functional anatomy of the 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 (figure 1).



Anterior view in flexion | Posterior view in extension

Figure 1 : Anatomy of the Knee Joint (Netter, 1987).

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 (figure 2).

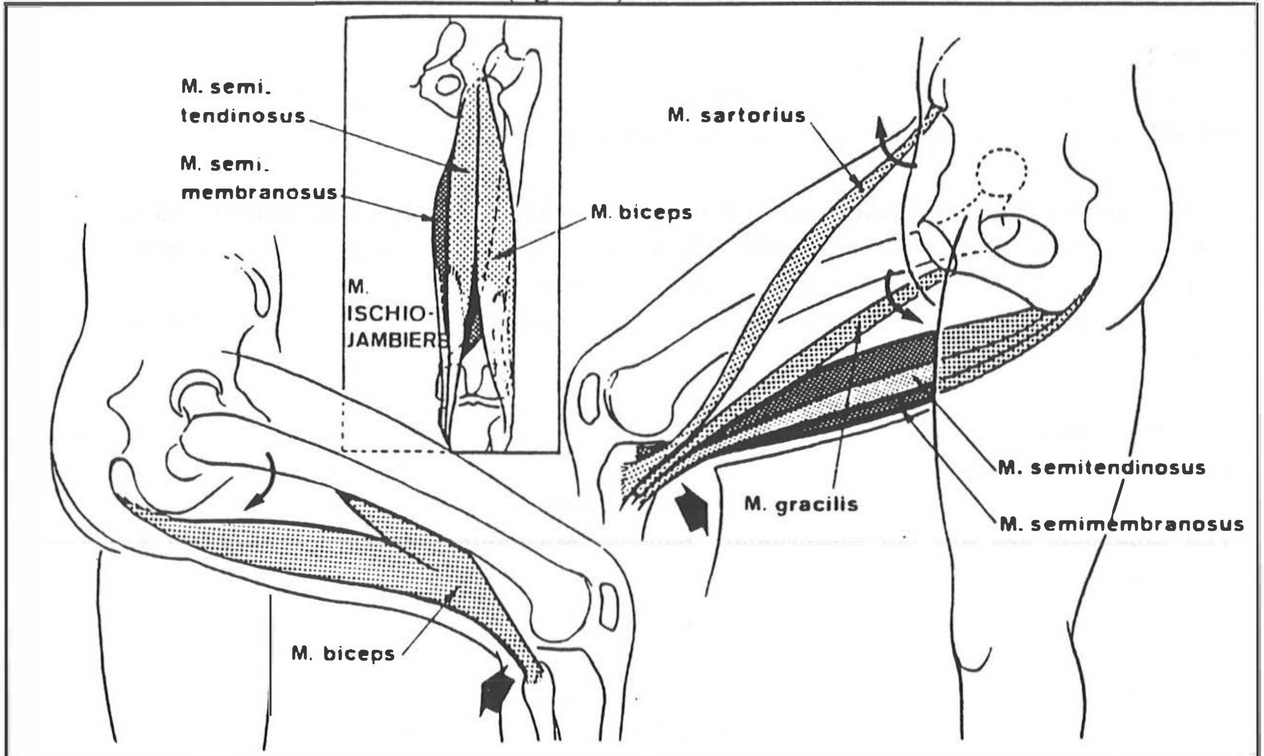


Figure 2 : Muscles around the knee-joint (Castaing et al, 1980).

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 knee injuries

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.

Knee joint model assumptions

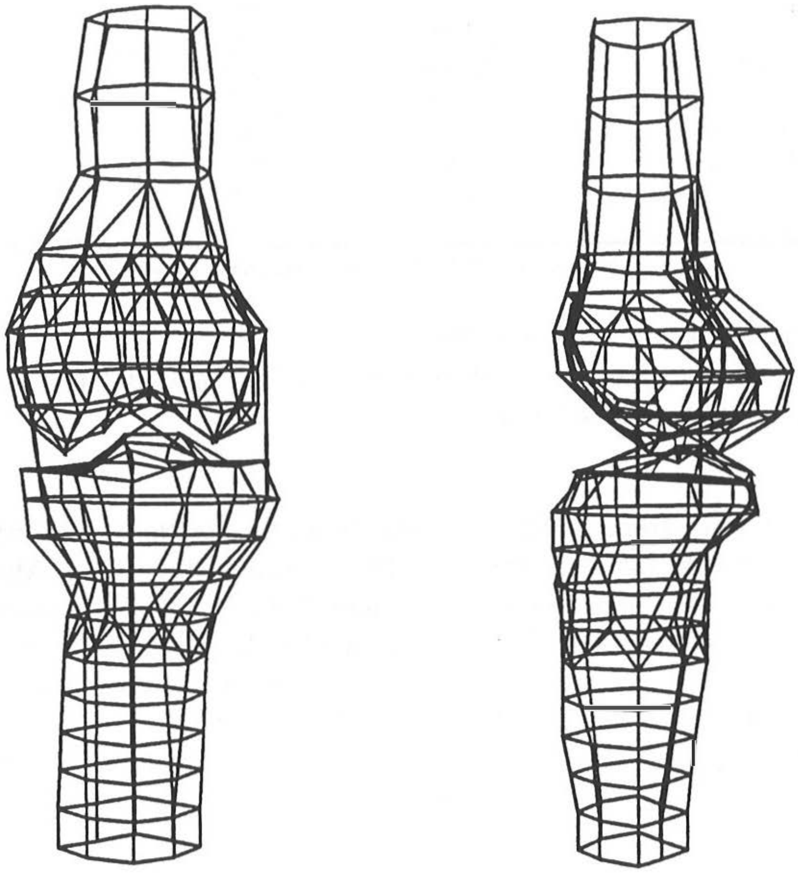
To analyse the complexity of the problem several assumptions are made to simulate the behaviour of the human knee joint in lateral impact :

- only the two bones are taken into account, the lower extremity of the femur and the upper extremity of 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.
- the posterior and anterior crucial ligament, and the medial and lateral collateral ligament are used.

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.

Methodology

The bones are described in 3d space. we reconstruct the shape of the femur and the tibia from human leg X-ray scanner. The geometry is discretized with shell elements.
the ligaments are described as a 1d linkage element (figure 3).



Anterior view

Side view

Figure 3 : Human knee model.

The material properties are found in the literature (Viano, 1986).

Mechanical properties of bones (Burstein et al, 1976) (table 1) :
The characteristic of the bone 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^3	1900	1900
Elastic modulus : N/m^2	$17.6\ 10^9$	$18.4\ 10^9$
Plastic modulus : N/m^2	$0.754\ 10^9$	$1.2\ 10^9$
Yield stress : N/m^2	$0.12\ 10^9$	$0.13\ 10^9$
Ultimate stress : N/m^2	$0.14\ 10^9$	$0.15\ 10^9$
Poisson's ratio : N/m^2	0.326	0.326

Table 1 : Mechanical properties of bones.

Mechanical properties of the ligament (Herzberg et al, 1981) (table 2) :
The main function of ligament is to resist tensile loading either due to muscular contraction or loads tending to displace the joint.
Mechanical properties of ligament have been determined by tensile testing of isolated tissues.

Ligament	Lateral collateral	Medial collateral	Anterior Crucial	Posterior Crucial
Elastic modulus : N/m^2	$1.9\ 10^6$	$0.8\ 10^6$	$0.6\ 10^6$	$0.3\ 10^6$
Ultimate stress : N/m^2	$40\ 10^6$	$20\ 10^6$	$20\ 10^6$	$10\ 10^6$
Rupture force : N	3000	3000	6000	6000
Ultimate Strain at rupture : %	30	40	60	60
Density : kg/m^3	1000	1000	1000	1000
Poisson's ratio	0.3	0.3	0.3	0.3

Table 2 : Mechanical properties of the ligaments.

The shape of the impactor is a half cylinder.
The geometry is discretized with shell elements. The material properties of the iron are chosen. The impactor weights 16 kg.

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 knee, 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 force applied laterally to the human knee joint (Kajzer, 1990). 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.

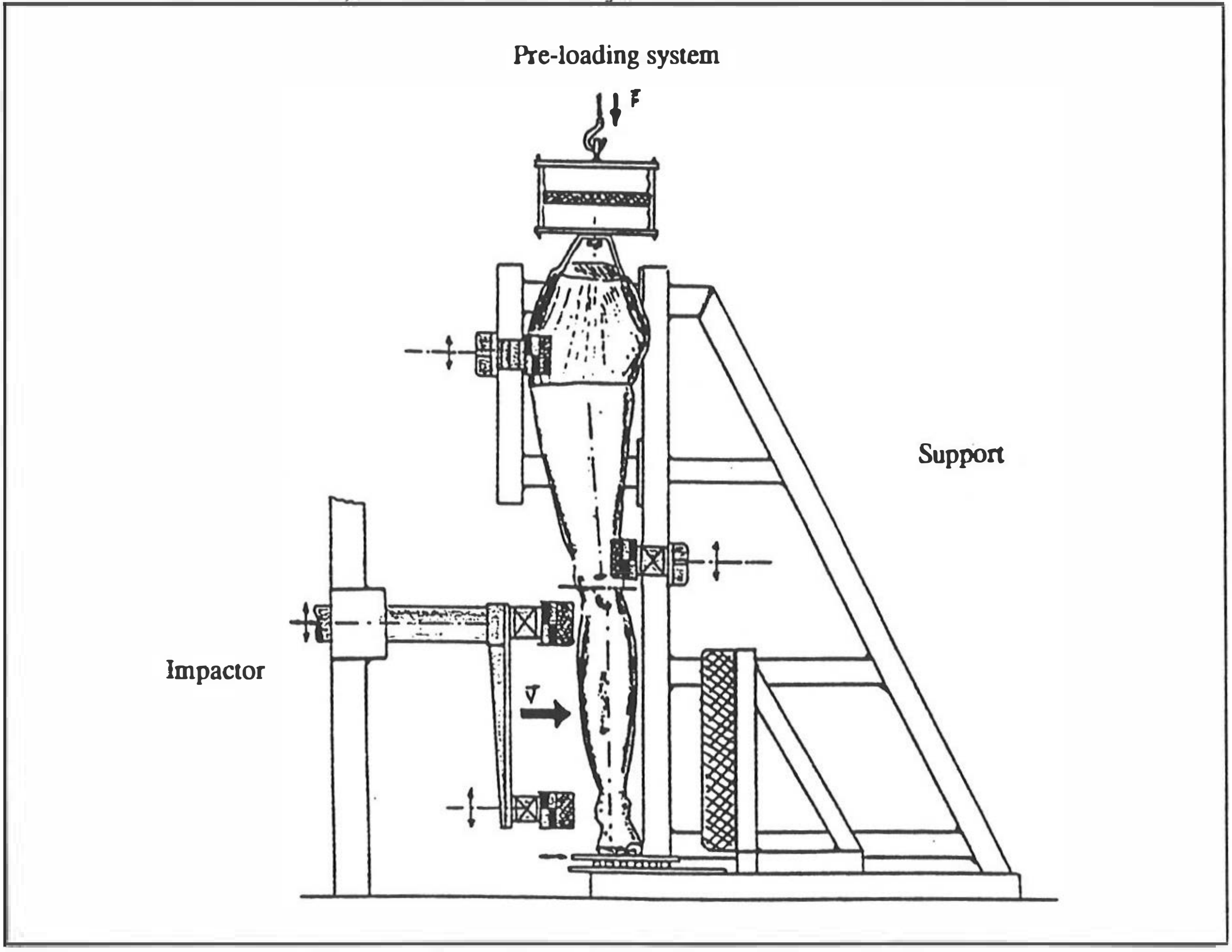


Figure 4 : Human leg in lateral impact (Kajzer, 1990).

19 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 propulsed by sandows. A specially designed impact arm equipped with force tranducers was mounted in front of the impactor. Figure 4 shows the initial of the tests procedure.

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.

To valited our model we compare the results from the tests with those from our model whitout forgotten our knee joint model assumptions.

Results

The results from the computer simulations are shown in figure 5 to 11, with the units from the System International.

The top and the external side of the tibia is impacted at 1.33 m/s.

The position at the first step and the deformed shapes at 8 ms, for the tibia impact test is presented at figure 5.

The maximum effective surface stress on figure 6 show the stress concentration around the area of the impact at 1.8 ms and near the insertion of the ligament at 3.8 ms.

On figure 7, the velocity magnitude and the acceleration magnitude curves for the node 180, on the top and on the anterior place of the tibia are filtered using SAE channel class 1000 Hz. There are peak values corresponding at the impact and after the curves oscillate due to the time step to large.

The figure 8 show the internal energy (due to the strain of the material), the kinetic energy (due to the displacement of the material) and the total energy, of the tibia and the ligaments. This curves are filtered using SAE channel class 1000 Hz. The ligaments accumulate more internal energy than kinetic energy, and the opposite appear for the bones.

The axial force in the ligaments are shown on figure 9 and are filtered using SAE channel class 1000 Hz. During the 14 ms the medial and lateral collateral ligament are more stressed than the anterior and posterior crucial ligament. After the axial force is the same for all the ligaments.

On figure 10 the maximum effective surface stress for the element 241 in front of the impactor present a peak value corresponding at the impact. This curve is filtered using SAE channel class 1000 Hz.

The bottom (of the mesh) and the external side of the tibia is impacted at 1.33 m/s.

The maximum effective surface stress on figure 11 show the stress concentration around the area of the impact at 0.2 ms. At 9.2 ms the stress concentration is located at the top of the tibia, where the condyle of the femur hits the tibia.

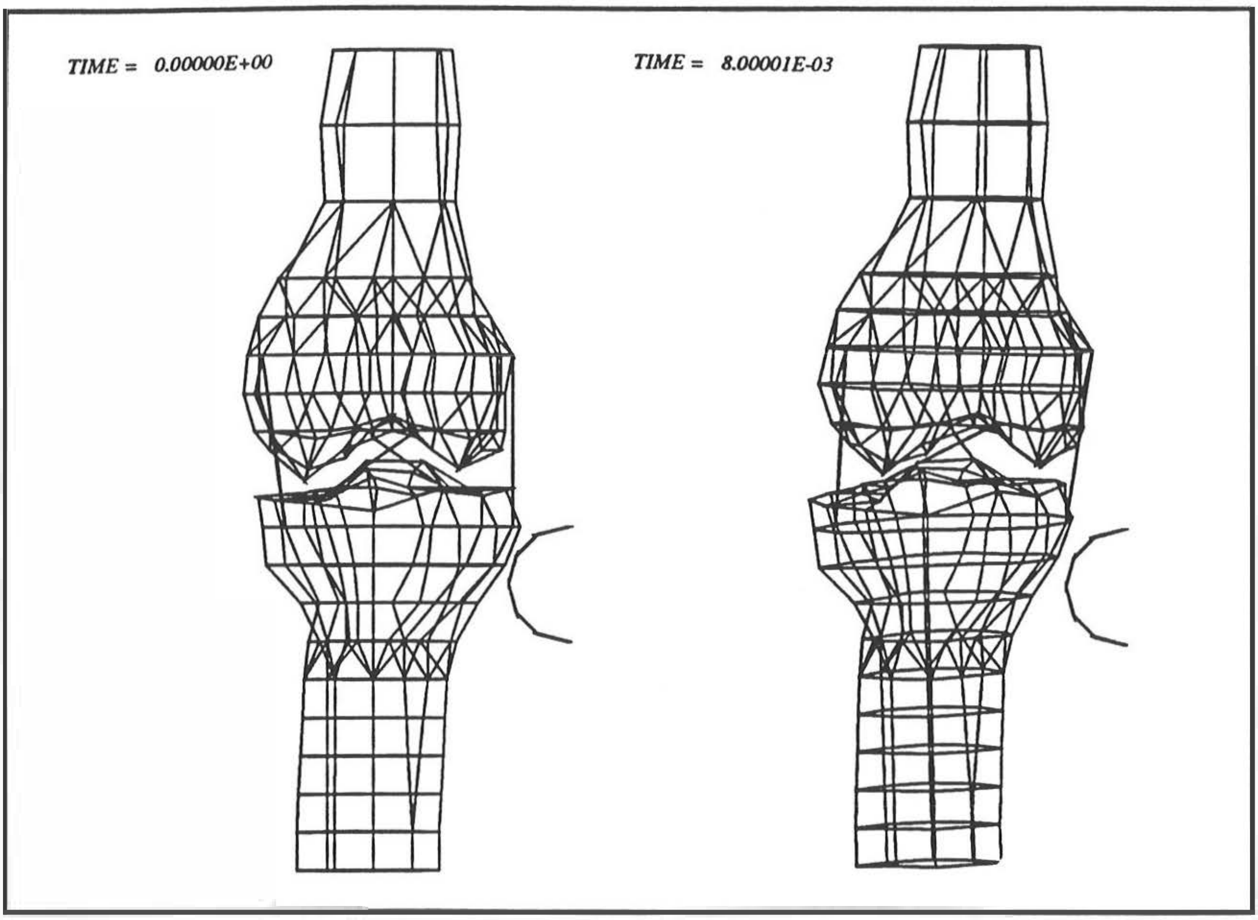


Figure 5 : Knee impact test. kinematics.

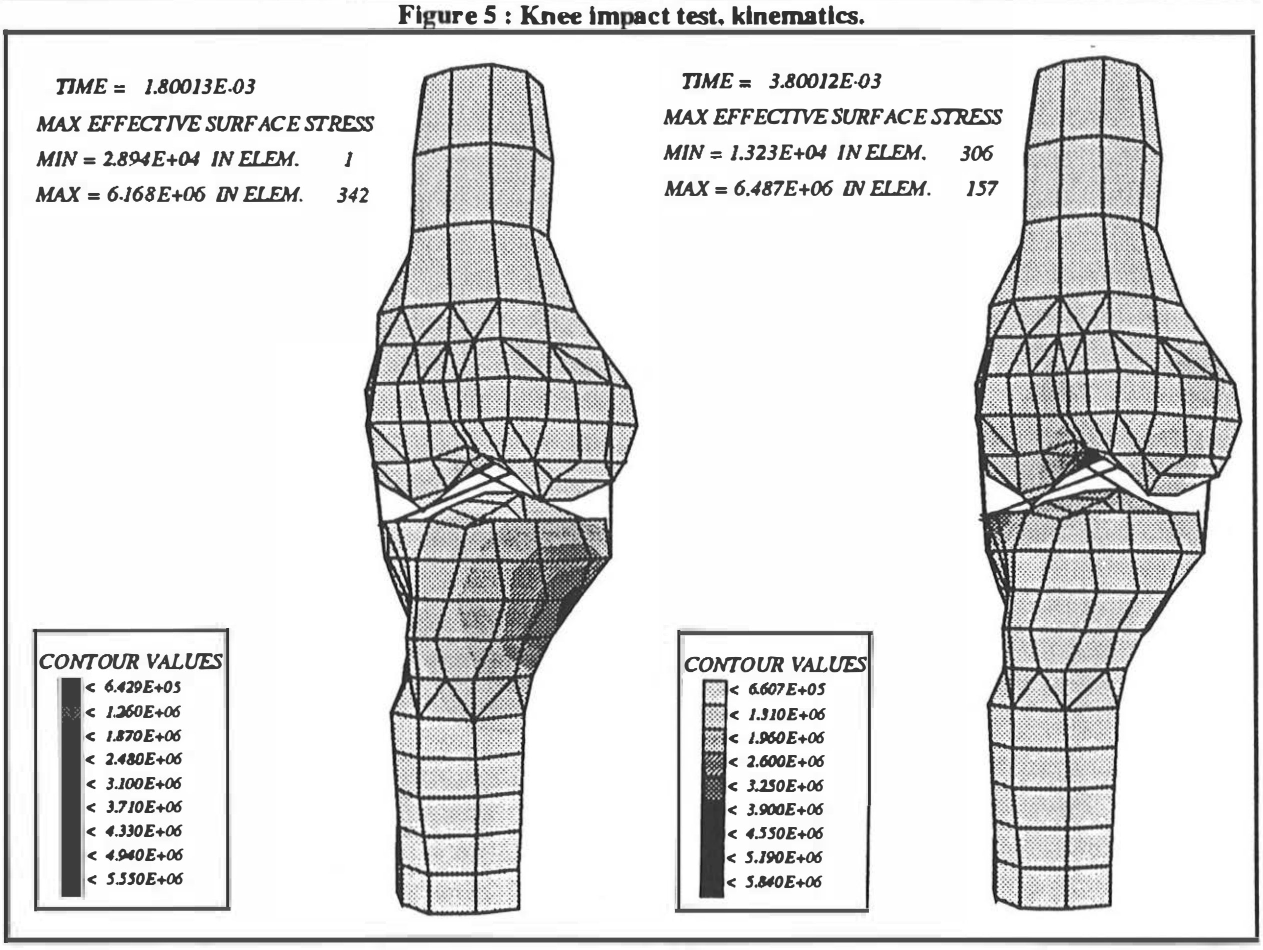


Figure 6 : Maximum effective surface stress.

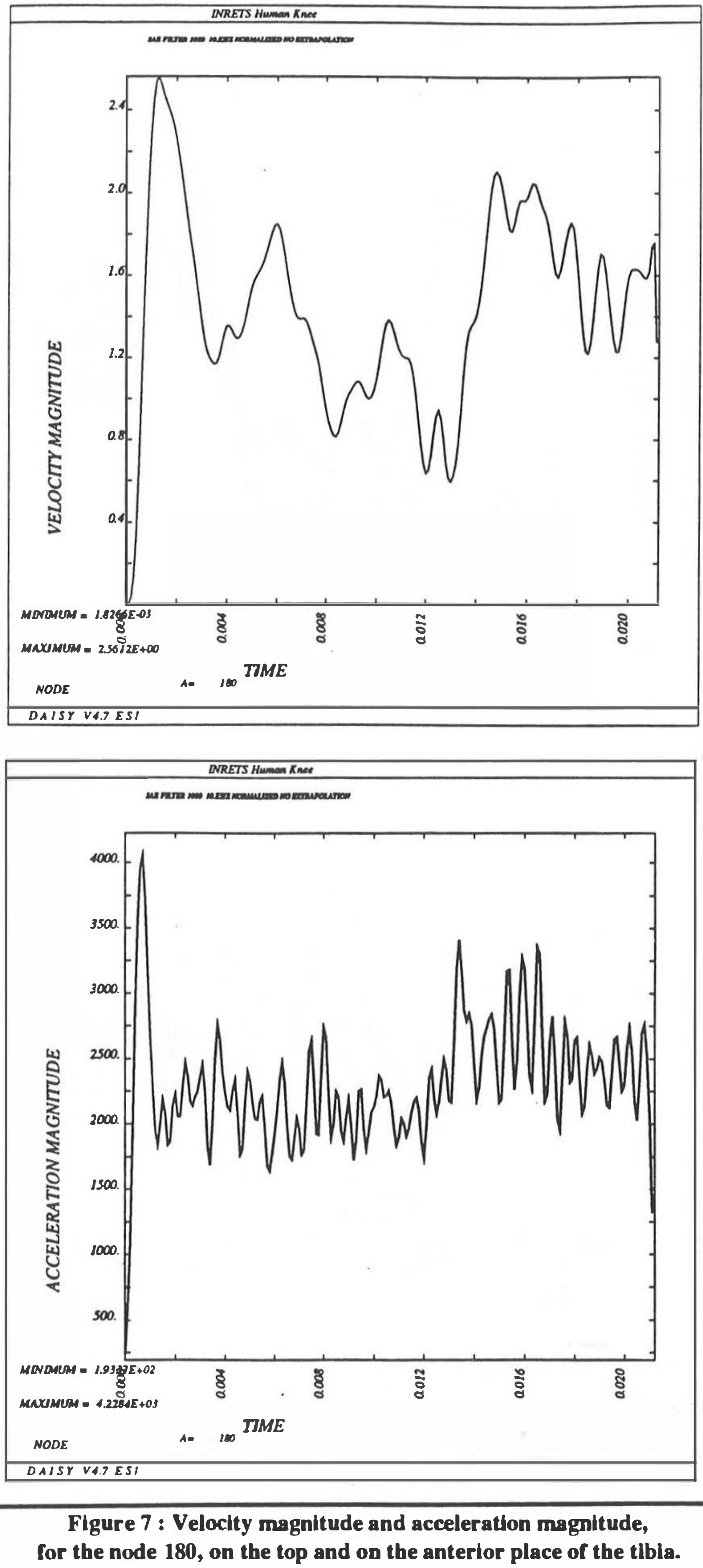
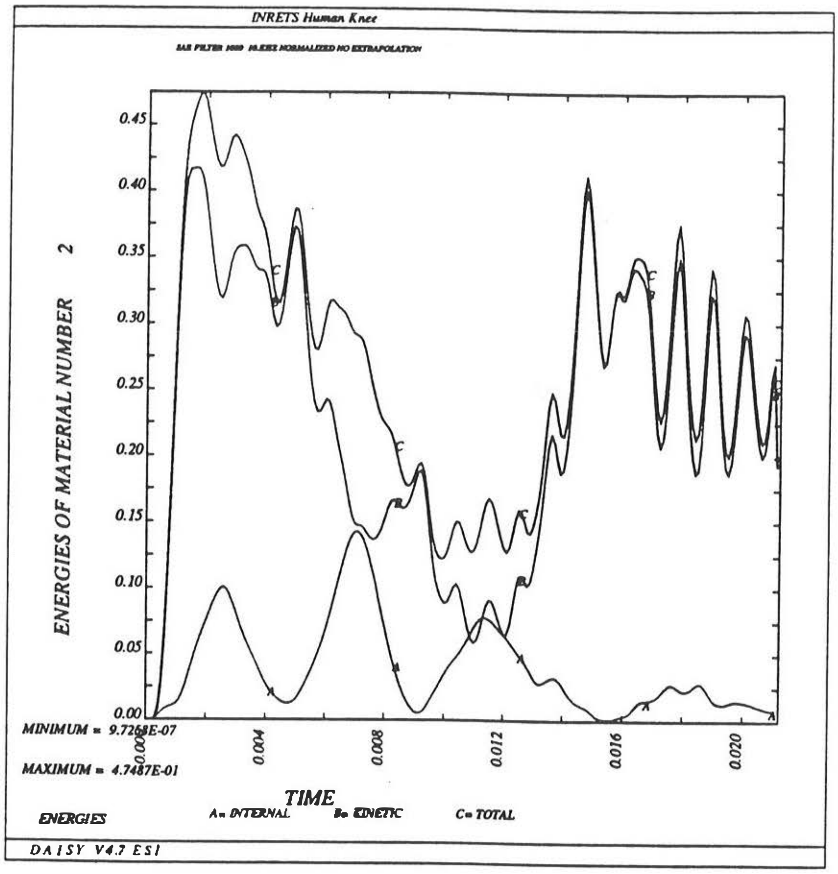


Figure 7 : Velocity magnitude and acceleration magnitude, for the node 180, on the top and on the anterior place of the tibia.



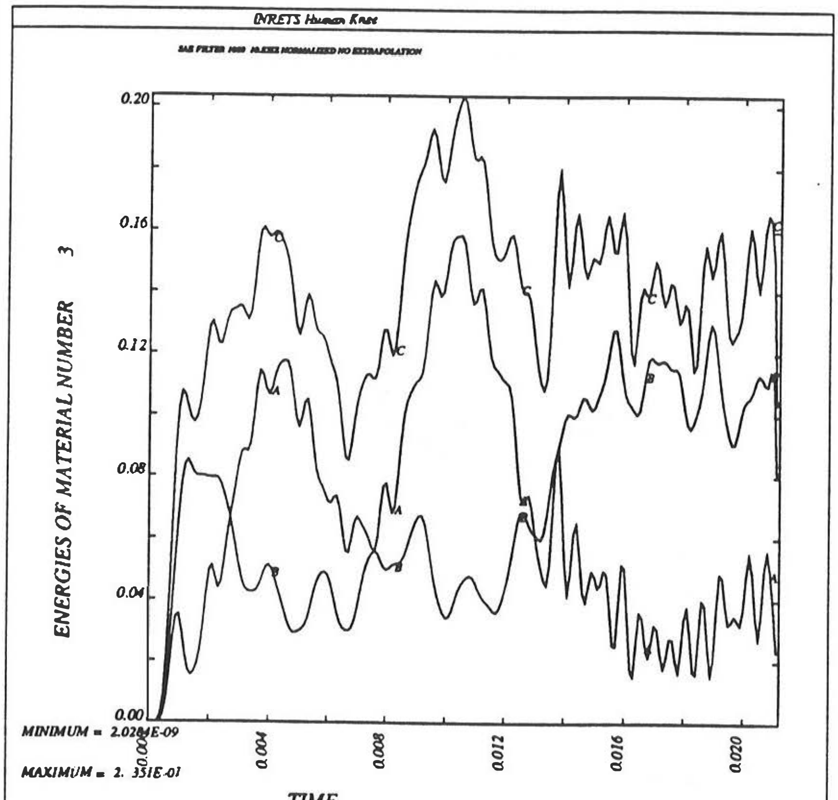


Figure 8 : Energies of material,
material number 2 : tibia, material number 3 : ligaments.

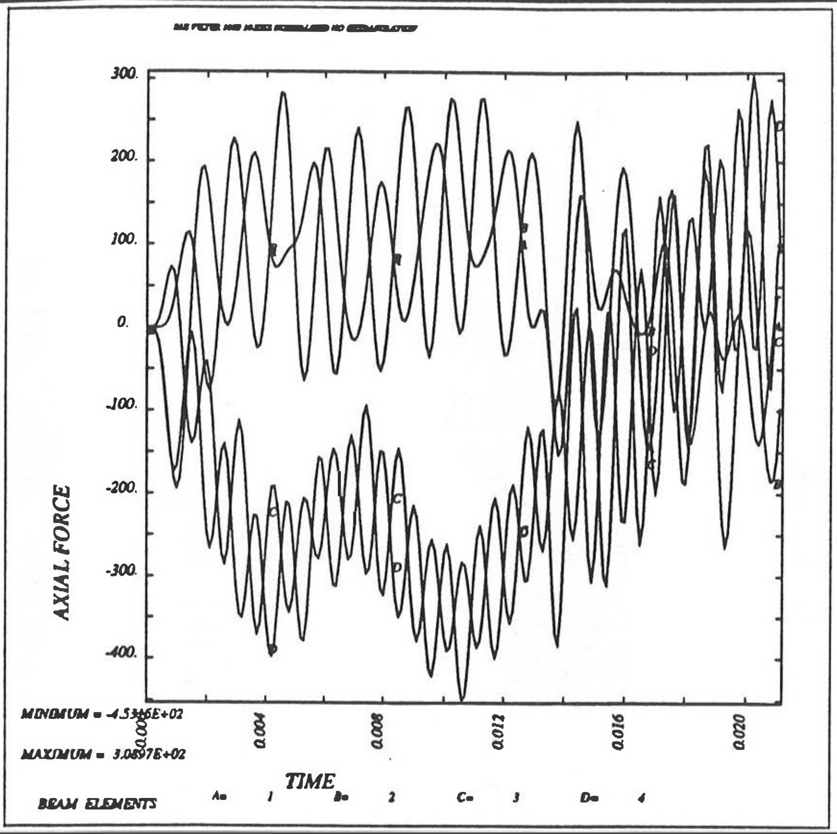


Figure 9 : Axial force in the ligament,
number 1 : medial collateral, number 2 : lateral collateral,
number 3 : anterior crucial, number 4 : posterior crucial.

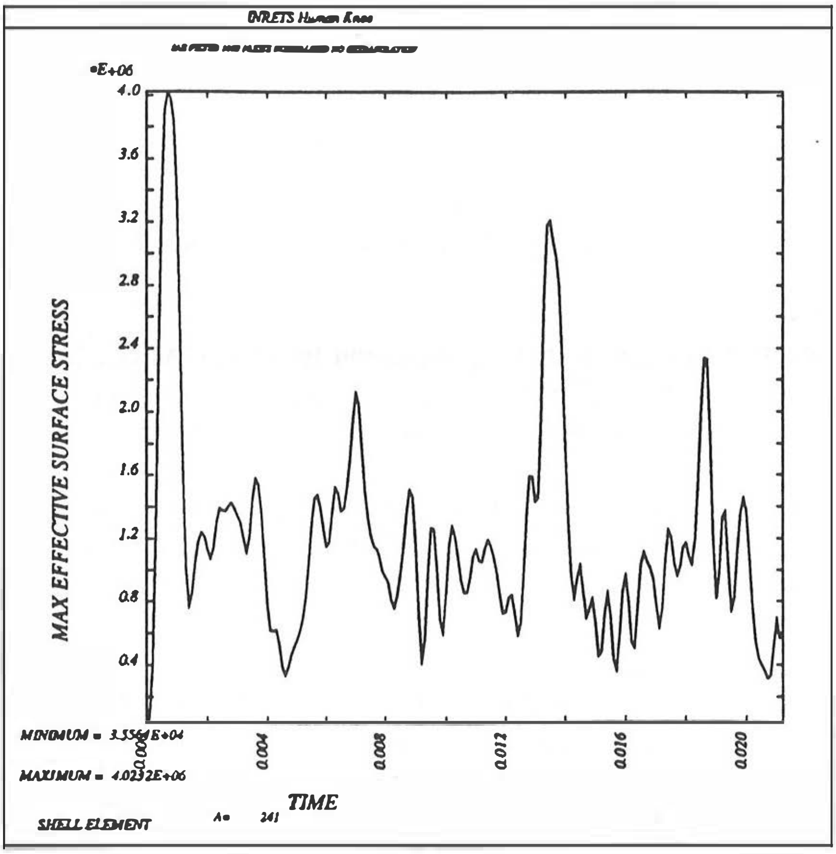


Figure 10 : Maximum effective surface stress,
for the element 241 in front of the impactor.

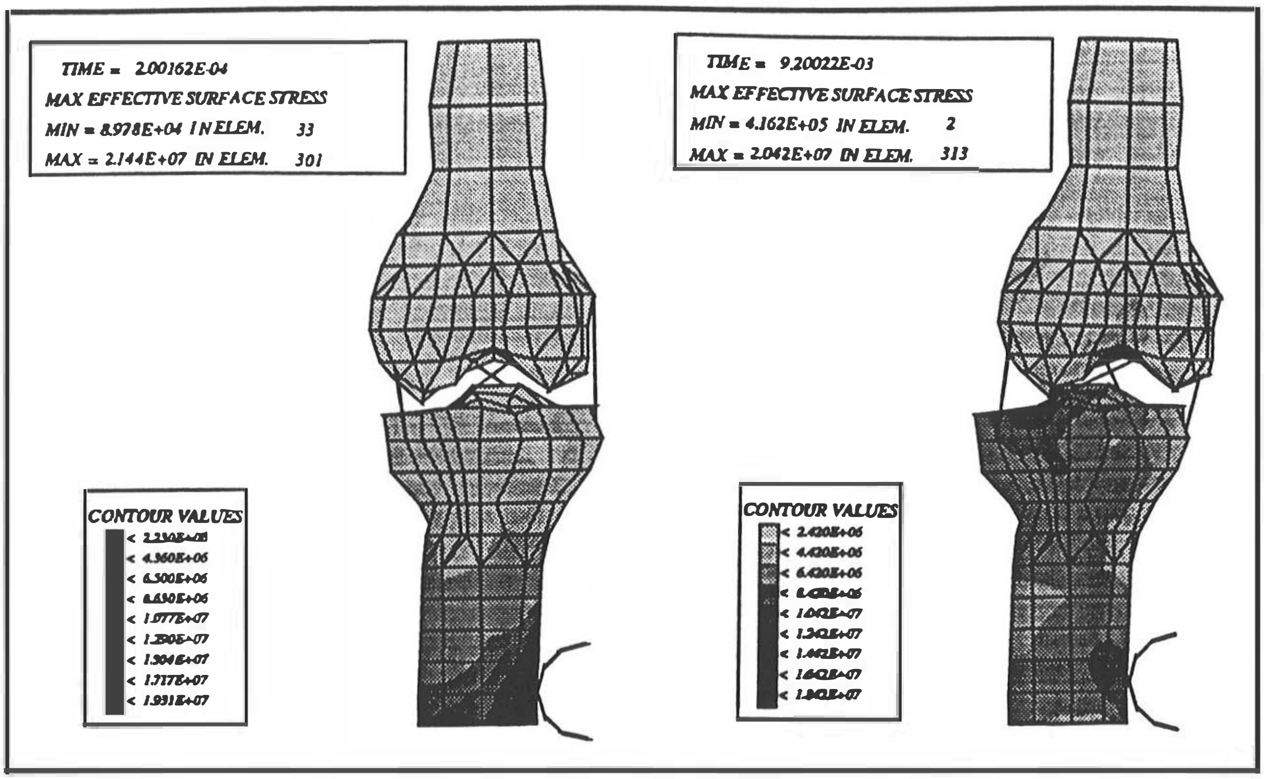


Figure 11 : Maximum effective surface stress.

Conclusions

The aim of this paper was to create a first step for a human knee model in lateral impact. This new approach with Finite Element Method gives information about the internal mechanical behaviour and also about the kinematics and displacement, velocity and acceleration.

Now the model is not 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.

Acknowledgement

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 and the X-ray scanner film of the leg was done by M. BOUALLEGUE, A. MORIN and J.P.H. NEIDHART du 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.
H. Burstein, T. Reilly, M. Martens. *Aging of bone tissue : mechanical properties.*
Journal of Bone and Joint Surgery, vol 58-A, 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.
EEVC/CEVE. *Pedestrian Injury Accidents.*
Proceedings of the Ninth International Technical Conference on Experimental Safety Vehicles, Kyoto, Japan, November 1-4 1982, pp 638-671.
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, 65, pp 113-118, 1981.
H. Ishikawa, K. Yamazaki, K. Ono, K. Sasaki. *Current Situation of Pedestrian Accidents and Research into Pedestrian Protection in Japan.*
Proceedings of the thirteenth International Technical Conference on Experimental Safety Vehicles, Paris, France, November 4-7, 1991.
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.
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.
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.
S.M. Sarin, Sharfuddin, Dindayal, M.L. Sharma, S.C. Kaushal. *Socio-economic Characteristics of Pedestrians and Cyclists Involved in Road Fatalities.*
Journal of Traffic Medicine, Vol. 18, No. 4, pp 279, 1990.
H.Vallée, C. Thomas, C. Tarrière. *Pedestrian Casualties : the Decreasing Trend.*
Proceedings of the Twelfth International Technical Conference on Experimental Safety Vehicles, Göteborg, Sweden, May 29 - June 1989, pp 1273-1278.
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.
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.