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Pierre-Jean Arnoux, Claude Cavallero, Patrick Chabrand, Christian Brunet. Knee ligament failure under dynamic loadings. *International Journal of Crashworthiness*, 2002, 7 (3), pp.255-268. 10.1533/cras.2002.0218 . hal-00166682

HAL Id: hal-00166682

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Submitted on 26 Nov 2017

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Knee ligament failure under dynamic loadings

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Abstract - A new experimental device was designed to study the failure properties of knee ligaments under dynamic loadings like those undergoing during pedestrian impacts. This device consist of a hydraulic actuator in order to impose various loading conditions on isolated knee ligaments. The originality of this device leads in the ligaments locking system obtained with spherical joint. Each sample is obtained with an isolated bone-ligament-bone complex embedded in resin. The sample can then be tested for various orientation of ligaments during loading. Results were obtained on twenty human ligaments subjected to tensile traction tests in their fibre axes at 2m/s velocity. Theses results shown that knee ligaments failure properties can be classified into two groups corresponding to the cruciate and the lateral ligaments. For each group it was also possible to identify specific damage pattern (with peeling failure and bone avulsion), failure mode (ductile and brittle for lateral and cruciate ligaments), damage localisation and deformation failure for both groups. These results were also used to validate a damage models for soft tissues structures.

INTRODUCTION

High-energy stress applied to body tissues during automobile accidents or competitive sports can lead to structural damage or fracture. To investigate the laws governing tissue behaviour under such stressful conditions tensile traction tests must be performed to measure tissue response. For the knee ligaments, high-velocity stress (similar to the forces applied during high-energy trauma) is applied to separate ligaments. A large portion of the data reported in the literature on the knee has been acquired with experiments designed to describe the forces acting on the different fibre bundles of the knee ligaments and tendons or to assess their viscoelastic properties under physiological loading. Many of the tensile traction tests performed have thus used quasi-static loading [1-6]. For cyclic loading, the strain rates used to achieve dynamic loading are commonly below $10^2\%s^{-1}$. These loading conditions used to validate viscoelastic models consist to apply forces with a strain rate lower than those occurring during dynamic loadings. In this last case, structure are submitted to very high strain rate level ($10^4\%s^{-1}$) for which the determined time necessary for a wave propagation in the whole structure is in the same range of value as those observed experimentally. Few works have been reported in this area [7,8].

Methods used to test the tensile strength of soft tissues can be divided into three categories, depending on the goal of the authors.

- The first method is to load an isolated ligament, both ends of which are blocked in resin with a moulding or chucking device [1,9-13]. This is not an easily reproducible method and the mounting procedure damages the ligament so the influence of ligament insertions cannot be studied.
- Another widely used method consists in fixing the distal part of the femur and the proximal part of the tibia in a mechanical system [2,3,14-19]. The ligament can also be set in specific directions (corresponding to different knee positions) before loading. Although this method can provide a considerable amount of data on the tensile properties of individual ligaments, there remains the problem of the contribution of the three other ligaments of each knee.
- A third method consists in using a bone-ligament-bone complex where the bones are embedded in a special resin (PMMA or epoxy) [5,6,20]. With this method, the load applied to the anterior and posterior fibre bundles constituting individual ligaments can be studied separately to examine the effect of insertion orientation on the mechanical properties and structural failure mechanisms.

Besides the difficulty in measuring dynamic tensile strength in such materials, data reported in the literature have shown wide variability, a common observation for living tissues. While such a wide

data dispersion is biological (“we are all different”), some parameters can be recognised as having a specific influence on tissue behaviour. Diament [21] and Noyes [16] showed that age strongly affects mechanical properties of human tissue with a subsequent effect on experimental results that Woo [3] suggested should be obtained for three age groups (22-35, 40-50, 60-90 yr). For strain-rate effects, two opposing hypotheses have been put forward: according to Fung [22] and Dorlot [23], ligament behaviour is strain-rate independent, whereas for Kennedy [24], Haut [25], Danto [26], Holden [27], and Pioletti [20] it is strain-rate dependent. Maes [28] and Lam [29] showed the effects of temperature and proposed a thermoelastic model for ligament microstructure. These authors noted that ligament collagen exhibits a physical reaction to high temperature (around 50°C), contracting before gradually desegregating. The orientation of the fibres during loading, also affects the mechanical properties observed with structure and failure models [5,32]. In order to obtain reproducible behaviour for cyclic loading, Fung [33] subjected samples to cyclic tensile tests before running experiments. Mass impact has been used in most of the dynamic loading experiments. The major drawback of such protocols used for dynamic and cyclic loading is that they do not allow quasi-static loading.

This study is a part of a global work performed on the human behaviour in car crash environments [35] and more specifically on the pedestrian injuries. During a pedestrian impact with vehicle impact velocities ranged between 20 to 40km/h, it has been shown that knee ligaments were subjected to high strain rate level and then severely injured. In such situation the injury mechanisms were isolated and then reproduced experimentally [37-38] with lateral shearing loading of the leg and bending loading around the knee. If it was possible to check with these tests the injury of the knee joints, no data were obtained on the failure properties of ligaments structures. In order to understand mechanical behaviour of ligaments under high speed loading and then their failure properties, a new experimental device was designed. This paper focuses on this device and also on results obtained in term of knowledge of failure properties of knee ligaments.

The device developed to achieve this experimental program meets three specifications:

- the study sample, composed of a bone-ligament-bone complex is embedded in resin
- both static and dynamic solicitation is achieved with a hydraulic actuator in order to study tensile properties of ligaments as well in static condition as with cyclic loading, as with dynamic loading up to failure.
- solicitations can be obtained for all realistic orientations of the ligaments (fibre direction, anatomical axis) in order to reproduce loading directions during accident (car occupant, pedestrian or sports injury).

The complete apparatus is presented below. Results are reported for 12 human cruciate ligaments and 8 human collateral ligaments subjected to tensile traction forces in the direction of the fibres axes at a velocity of 2m/s.

MATERIALS AND METHODS

Experimental apparatus

The device (Fig. 1) consisted of an actuator set on a mobile beam in the z direction allowing application of an initial force independently of the actuator. Samples were attached to the device by two sets of orientation and locking systems: one linked to the actuator and other coupled to a cross table. A hydraulic actuator (Schenck PLF 7D) produced the solicitation. Harmonic solicitation (with frequencies up to 1 kHz) can be used to precondition the ligament. For dynamic solicitation, the actuator had a nominal 20-mm stroke and a maximum force of 5.2kN. The cross table allowed 2 degrees of freedom in the (x,y) plane for the lower end of the sample. Both ends of the sample (i. e. one coupled to the actuator and the other to the table) could be rotated around the z axis. In order to obtain a precise locked orientation of the ligament during solicitation, bone insertions were embedded in a spherical mass of polyurethane resin. A spherical joint gave each ligament insertion 3 degrees of freedom (degrees of rotation around the x, y and z axes). Each orientation and locking system was

made of two parts (a main part and a cover) in which the spherical mass was placed (Fig. 2 and 3). A conical hole allowed inserting a “joystick” into the spherical end of the sample to control ligament orientation before locking (Fig. 3.a and 3.b). Force measurements were recorded with a Kistler 9067 A4 tri-dimensional force transducer for which maximum forces applied in the x (or y) and z directions were 5kN and 10kN respectively. Displacement measurements were made with a specific sensor inside the actuator. Acceleration was determined with 200g piezo-resistive sensor (EGE) pasted using cyanolite stick on the mobile parts i. e. on the fixing sample system which was between the ligaments and the actuator (cf. figure 1 and 2). Measurements were obtained with a 10kHz sampling frequency for 5.8s.

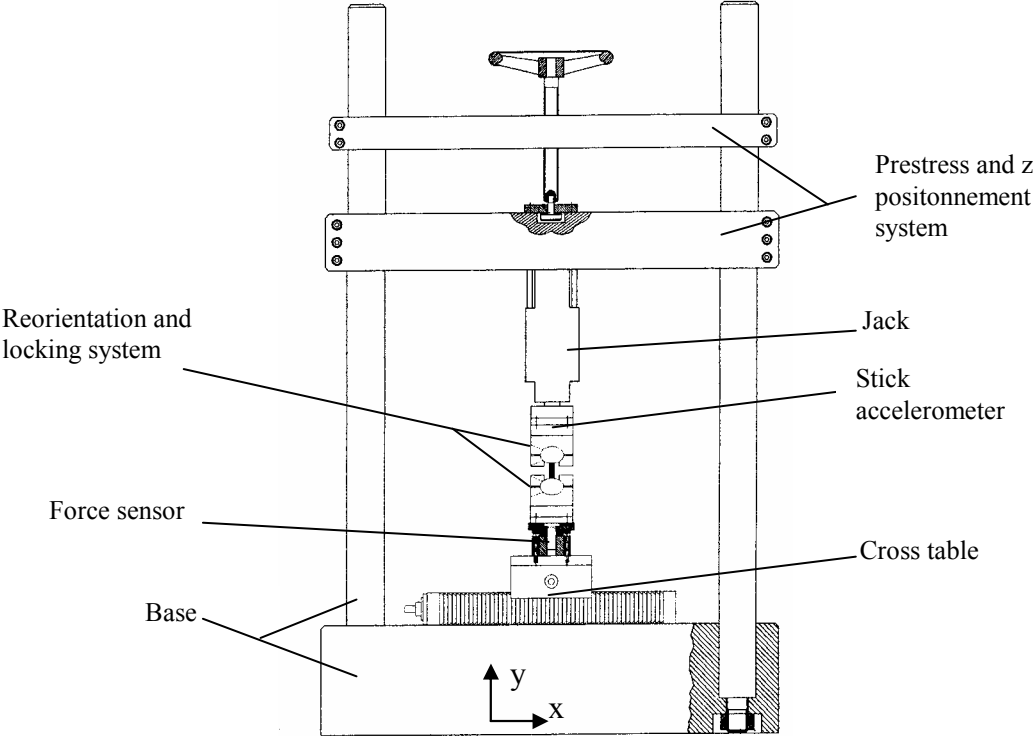


Figure 1: The complete Apparatus

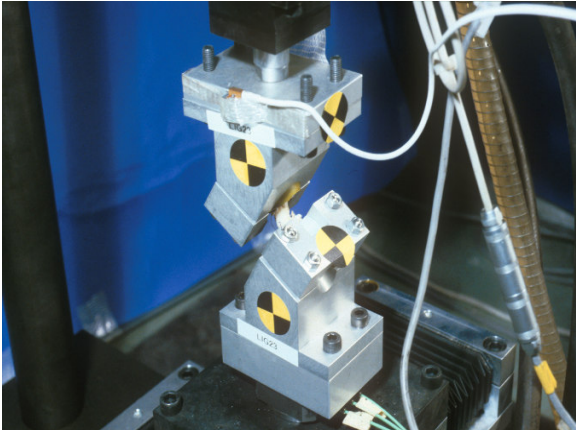


Figure 2: The orientation and locking system

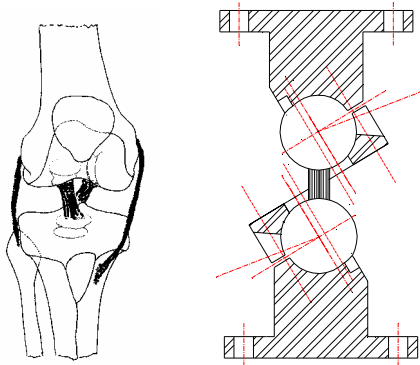


Figure 3.a : Mounting device for cruciate ligaments

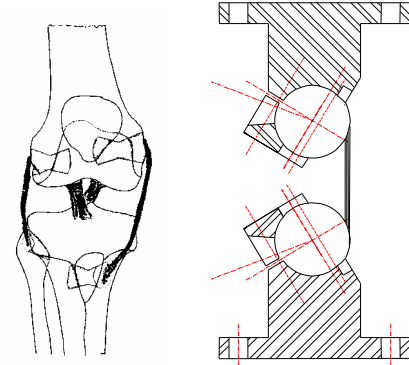


Figure 3.b : Mounting device for lateral ligaments

Specimen preparation

Knee ligaments from 4 human cadavers (an 88-year-old man and 3 women aged 58, 81 and 89 years) were tested. The cadavers (obtained from the Marseilles University Faculty of Medicine) were treated with Winckler solution [34] to ensure proper preservation of soft tissues, then held at +3°C until testing. Prior to testing, the knees were thawed to room temperature for 3 hours, then x-rayed to check bone integrity. The knees were then dissected and all skin, muscle tissue, and joint capsules, as well as the patella and tendons were removed, leaving the proximal part of the fibula and tibia, the distal part of femur, the menisci and the four knee ligaments. Before extracting the bone-ligament-bone complex, the articulation was locked in full extension with a Hauffmann external fixator using a pin in the distal part of the femur and the proximal part of the tibia. The four bone-ligament-bone complexes were then dissected (cf. figure 4). The collateral ligament insertions were dissected following the axes of the lower limb. The cruciate ligaments with the corresponding bone insertions were dissected on the tibial spinal eminence and between the two femoral condyles (cf. figure 4). The dimensions of each ligament and the angle it formed with the insertion surfaces were recorded before and after removing. In accordance with work by Noyes and Grood [16], the ligament was assumed to have an elliptic cross section. The two axes of this elliptic section were determined at the tibia insertion, at the medial portion of the ligament and at the femoral insertion. The length of the ligament was calculated from three measurements taken by placing the calliper on the ligament insertions. Each bone insertion was then embedded in a truncated spherical mass of resin measuring about 30 mm in diameter. The mechanical properties of the two-component polyurethane resin (F1-Excel-Euroresin) were sufficient to ensure good fixation of the bone insertion and to support the dynamic loading applied during the tests. Complete resin polymerisation was achieved at ambient temperature in a few hours at a exothermic maximal temperature below human body temperature (which was checked using temperature sensor). After terminating sample preparation, the entire sample was plunged into a Winckler solution until tensile testing.

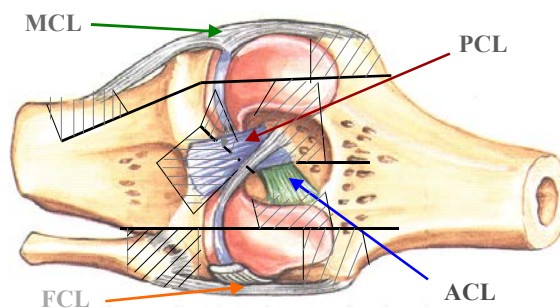


Figure 4 : dissecting plan for the bone ligament bone complex with a posterior view of the knee.

Experimental procedure

Prior to each test, the bone-ligament-bone complex was placed in the orientation and locking system before being mounted on the complete apparatus. An initial force between 5N and 30N was then applied and the spatial positions of the two systems were determined. All tests were performed in the direction of the fibre axes. A constant velocity 1.95m/s load was applied to one end of the ligament, the other end being held in the fixed position. The chosen velocity was based on earlier studies on automobile-pedestrian impacts (with velocities ranging between 25 and 39 km/h) performed in the laboratory [35] which showed that the corresponding velocity for each ligament was approximately 2m/s. Under these conditions, loading time was less than 1 ms, which can be considered a Dirac impulse and was checked for each test by the superposition of acceleration, displacement and force curve (cf. figure 5). The corresponding acceleration was around 120 g. No preconditioning of ligaments was used due to the non harmonic solicitations used in this study. All tests were run at room temperature (23°C). In order to keep the ligament sufficiently moist, Winckler solution was regularly applied on the ligament with a compress. The whole experiment was filmed with a high speed camera (1000 pictures per second). The data were digitalised and filtered by a channel class 1000Hz filter.

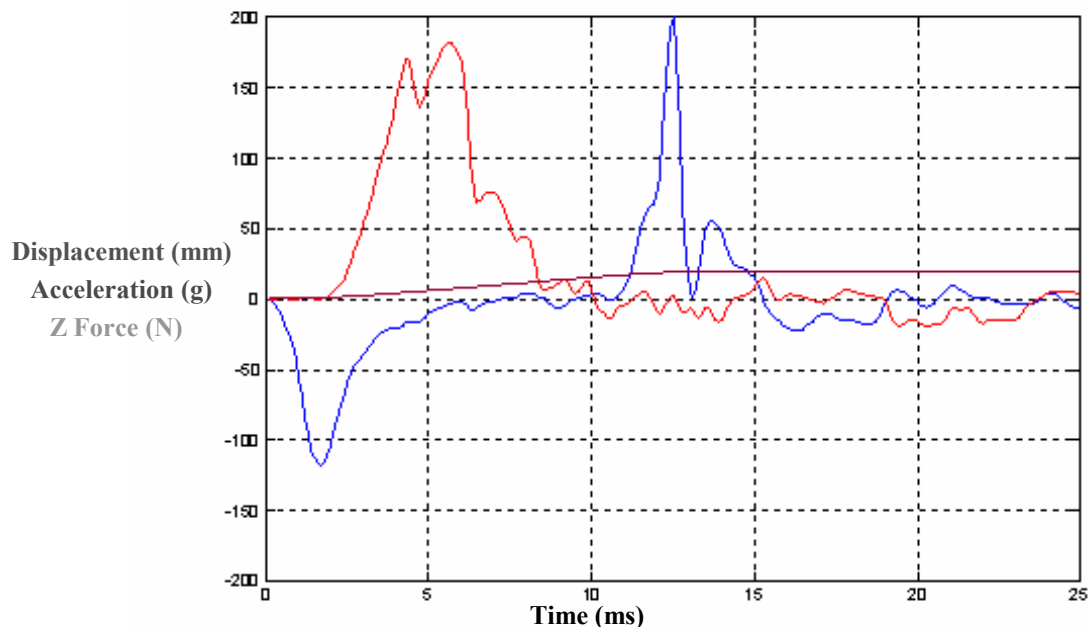


Figure 5: superposition of acceleration, displacement and force curve in the case of an anterior cruciate ligaments

RESULTS

For all tensile tests, the forces in the (x,y) plane were negligible compared to those in the z axis, confirming the correct orientation of the ligament during solicitation. In the subsequent force versus displacement curves (Fig. 6.a, 6.b, 9.a and 9.b), the zero force at test onset was translated to the initial force value (i. e. initial pre-load minus ligament relaxation before the tensile test).

For the anterior (ACL) and posterior (PCL) cruciate ligaments

As was showed by Fung [33] and Vidiick [11], the force-time curve (Fig. 6.a & 6.b) exhibited three phases:

- In the first non-linear phase, the force increased exponentially with time, describing the physiological range of solicitation. From a mechanical point of view, this first step corresponded to stretching of the two main fibre bundles, which have a wave shape in their relaxed structure. This first phase was strongly dependent on the initial force applied to the structure. When applying an initial

force greater than 20N, the slope of the force versus displacement curve showed that the ligaments rapidly reached the linear elastic phase. Since the initial pre-load was between 5 and 20N and the strain-rate level was very high, this non linear elastic phase was not as prominent as observed in low strain level tensile testing [20,4,33]. In soft tissues, the toe region generally produces a considerable amount of structural deformation. This was not the case in this study. The explanation could be the initial pre-load, but for ligaments with low pre-stress (Lig12, Fig 6.a) results were equivalent. As fibres are recruited in the structure, stretching and alignment generally increases the pressure on the amorphous substance, causing fluid exudation, interpreted as a viscous component of the tissue behaviour. With high strain-rate levels, fluid exudation was reduced by fibre tensile effects [30-31]. When the fibres had stretched to their maximum, the structure moved from the non-linear to a more linear behaviour.

- The second phase was fairly elastic. The stiffness and then the Young modulus of the structure were calculated. The stiffness of the structure ranged between 50 and 145 N/mm for the anterior cruciate ligaments and between 60 and 180N/m for the posterior cruciate ligaments. According to the assumption made on ligament cross section and measurement accuracy, the corresponding Young Modulus were between 25 and 210MPa and between 11 and 51 MPa for the cruciate and collateral ligaments respectively. These values are in agreement with those found in the literature [3,16,17,36, 39,40] but are in the lower range proposed by the last authors. These low values could be related to the age of the cadavers, to the preservation method or to strain-rate effects. High Young modulus values, known to result from more rapid strain rates, were not observed in this study. As a strong initial force was applied to the structure (in order to limit the toe region), viscous effects and then strain-rate effects may not have been significant. The age of the subjects, their gender and activity before death were certainly different and also could have affected the present results. Finally, the preservation method could have affected the soft tissues properties, but no study has been conducted to compare the tensile properties of fresh and Winckler-preserved soft tissues.

- The final phase was a region of brittle damage leading to failure. Two peaks were observed. These peaks corresponded to the respective failures of the two main fibre bundles of the structure (Fig.6.a and 6.b). Each cruciate ligament being constituted by two fibre bundles with different length and orientation, loading was not identical, leading to a two-step failure pattern. The ultimate force at failure differed from one test to another, but the ultimate deformation always occurred between 18 and 24% (according to the initial force for both the anterior and posterior cruciate ligaments). All cruciate ligaments tore off the tibia insertion. The failure pattern was not the same for the two bundles of fibres. The anterior bundle peeled off its insertion, while the posterior one pulled out of the thickness of the cortical bone (Fig. 7.a, 7.b, 7.c and 7.d). The observed behaviour (Fig. 6.a and 6.b) was associated with the failure peaks of the two fibres bundles. In light of bone behaviour [41], bone avulsion could correspond to the more brittle peak in the force-time curve whereas peelings would correspond to the more ductile failure peak. Therefore, while failure was observed at the ligament insertions, the entire structure was assumed to have been damaged since swelling effects were observed after failure (Fig. 8). These failure patterns were in total agreement with the results of impact tests observed at complete knee extension (bending or shearing tests) [42].

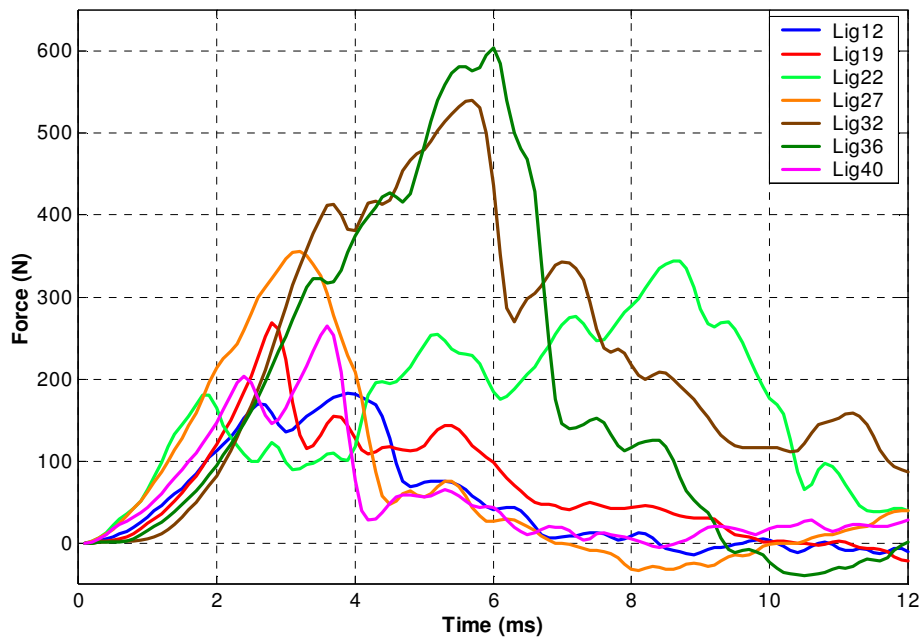


Figure 6.a: forces versus time curves for the anterior cruciate ligaments

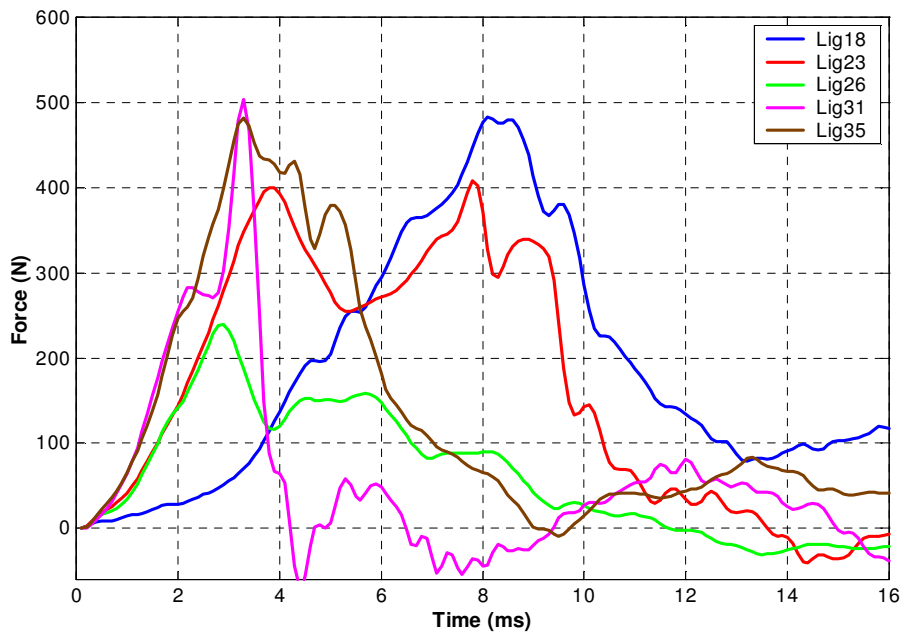


Figure 6.b: forces versus time curves for the posterior cruciate ligaments

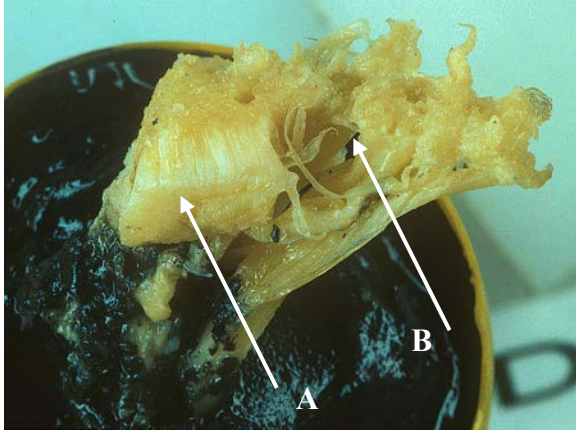


Figure 7.a : ACL failure profile. A was for the bone avulsion and B was for the peeling failure

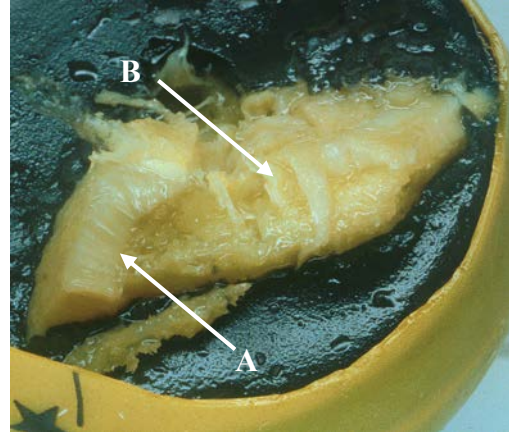


Figure 7.b : ACL failure profile on the bone insertion A was for the bone avulsion and B was for the peeling failure

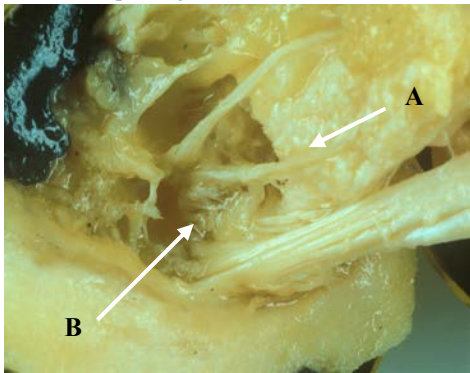


Figure 7.c : PCL failure profile. A was for the bone avulsion and B was for the peeling failure

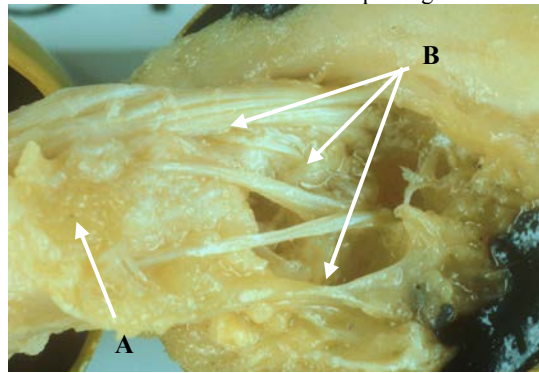


Figure 7.d : PCL failure profile. A was for the bone avulsion and B was for the peeling failure

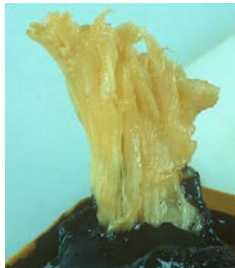


Figure 8: Swelling profile of the structure

For the medial (MCL) and fibular (FCL) collateral ligaments

As observed for the cruciate ligaments, the force-time curve of the collateral ligaments could be divided into in three phases: a stretching phase, a quasi-linear elastic phase and a damage phase (Fig. 9.a and 9.b).

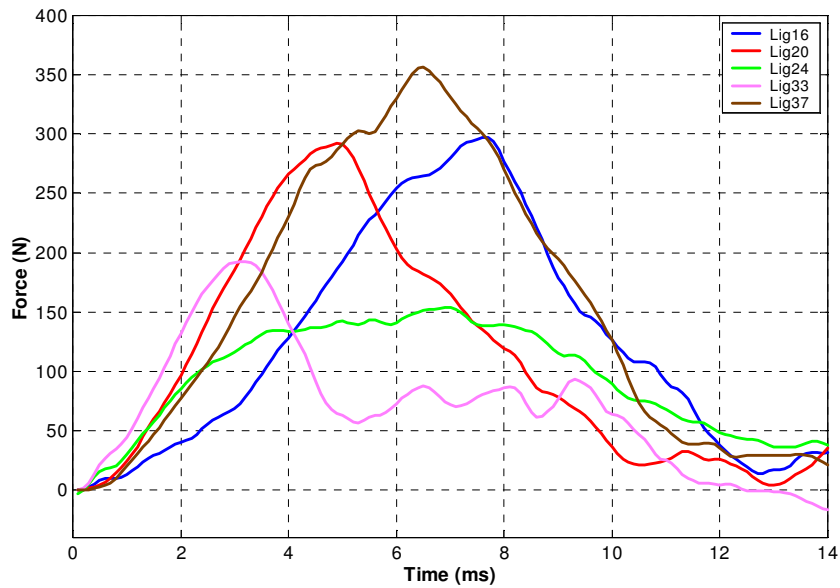


Figure 9.a: forces versus time curves for the medial collateral ligament

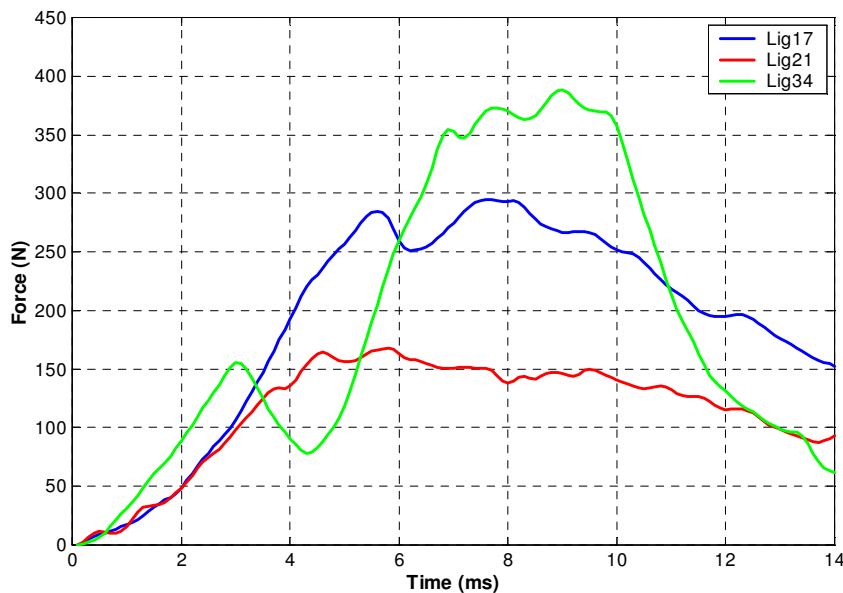


Figure 9.b: forces versus time curves for the tibial collateral ligament

- Despite the initial force applied to the collateral ligaments, the toe region was clearly observed for initial forces between 9 and 28N (Fig. 9.a and 9.b).

- The elastic phase stiffness was between 33 and 53N/mm for the MCL and between 28 and 78 for the FCL, which was less than observed for the cruciate ligaments. The corresponding Young modulus (according to the assumed cross section and measurement accuracy) were ranged from 12 to 51 MPa and 19 to 32 MPa for the MCL and FCL respectively.

- Since collateral ligaments are composed of only one fibre bundle, the damage phase described only one peak on the force-time curve. Compared with the cruciate ligaments, this was a very ductile peak. Depending on the initial force applied to the structure, deformation failure occurred at 25 to 38%. The failure occurred at the femoral insertion in more than 80% of the samples and exhibited a bone avulsion pattern (Fig. 10.a and 10.b). As cortical bone thickness, and thus resistance, at the insertion of the collateral ligaments was very thin, the failure mode (probably in mode I) was not comparable with the failure modes of the cruciate ligaments (combined mode I and mode II). This observation

demonstrated that the structural properties of the ligament insertions are not the same for the cruciate and collateral ligaments. Assuming the insertion has a ligament-to-bone hydroxyapatite gradient, this gradient exhibited a strong discontinuity for the collateral ligaments. In addition, the ligament-bone interface was very thin and can be considered as a membrane which was pulled out during the damage phase. We also observed damage throughout the structure, with swelling and peeling effects.

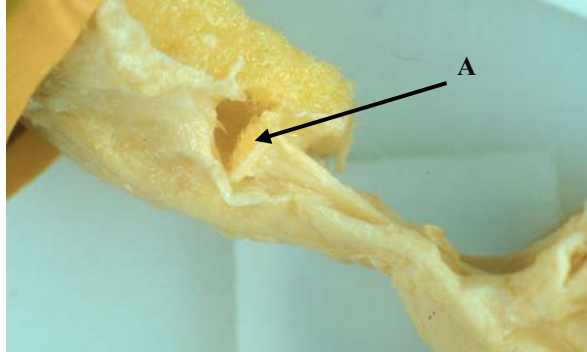


Figure 10.a : MCL failure profile. A was for the pull out for the interface between ligament and bone

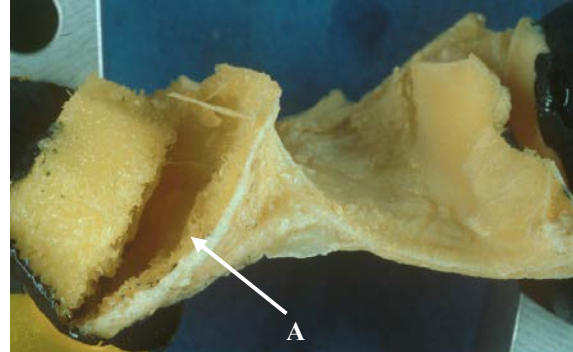


Figure 10.b : TCL failure profile. A was for the pull out for the interface between ligament and bone

DISCUSSION

Our new experimental device allowed dynamic tensile testing on a bone-ligament-bone complex for all possible loading orientations. Completing information on the mechanical properties of the knee ligaments exposed to dynamic stress, our experimental results demonstrated their failure behaviour when subjected to tensile loading in the direction of the fibre axis.

Experimental apparatus

Although the two cruciate ligaments are closely neighbouring structures, the experimental apparatus enables study of each ligament separately without affecting its structure or insertions. With the Hoffman fixator, the joint is held in a stable position, with no deterioration in the accuracy of the measurements of the relative orientations of the ligament insertions. The relative orientation of the knee ligaments, from dissection through sample fixation on the orientation and locking system, will soon be controllable with a 3D acquisition system (“faro arm”). Whatever the method used, accurate biometry is very difficult because of the shape and size of the bone insertions. We used the Noyes & Grood method that gives an accuracy of about 2mm. A more recent method [43,19] could be used to increase the accuracy of section and length measurements. Polyurethane resin is very efficient with such fat and wet material because of the short polymerisation time at a temperature under 38°C. Nevertheless bone insertions had to be “anchored” to ensure proper moulding. Moulding of fibular collateral ligaments required particular attention due to the tibiofibular articulation. Together with the reorientation and locking system, the spherical shape of the sample extremity was most efficient, allowing (with the spherical joint) a precise orientation of the ligaments, both for cruciate and collateral ligaments. Compared with other dynamic loading systems, the hydraulic actuator ensured a very precise control of the loading force and helped avoid measurement noise. Other loading modes (static, dynamic harmonic, relaxation...) can also be achieved. Since ligaments are naturally pre-loaded, it was necessary to obtain a good alignment of fibres before tensile testing (in order to demonstrate the damage phase up to failure). As the nominal stroke of the device was limited (20mm maximum) we decided to induce the initial force independently of the actuator (using the mobile beam to which the actuator was coupled). The failure processes which occur for an elongation of less than 10mm were not affected by this limited stroke. Measurements made with the 3D force transducer were taken separately for the initial force and during loading. Consequently, the ligament structures had time to relax between application of the initial force and the test time. The 0N level on the time

curve at the beginning of each tensile test therefore corresponded to the value of the initial force (with an unknown relaxation factor). This error in results was estimated to be less than 2% of the real force.

Experimental results

Although we used very high strain rates, our results for all ligaments studied were qualitatively the same as those presented in the literature. A quantitative comparison with data reported earlier is not exactly appropriate because of the preservation method used. Further experiments will be needed for such a comparison. We found that the force-time relationship could be then divided into three phases, a toe phase for the structure, a linear elastic phase and a damage phase up to complete failure of the structure. Since the cruciate ligaments are made of two bundles of fibres, which are not oriented in the same direction, the failure pattern exhibited two peaks on the force-time curve. These peaks can be attributed to the separate failures of the two fibre bundles.

With pre-loading we observed a relaxation between application of the initial forces and the beginning of tensile testing. The different pre-loading had no effect on failure behaviour. Therefore, by minimising viscous effects, high pre-stress might dampen strain-rate effects.

The accuracy of the structure's Young modulus is highly dependent on the accuracy of the ligament section measurement. It is therefore more relevant to evaluate the (1D) stiffness of the structure. The stiffness measurements we obtained showed a wide spread and were less than those presented by others authors [3,4,19]. These differences might have been induced by strain rate, the cadaver preservation method, or the age and sex groups of the study population (3/4 were old women) and more generally to the variability of living tissues.

The failure occurred rapidly for all ligaments tested. While the ultimate force at failure showed a strong dispersion (between 170 and 500N for the cruciate ligaments and between 150 and 300N for the collateral ligaments), the corresponding ultimate deformations (according to the initial force applied) were between 18 and 24% for the cruciate ligaments and between 25 and 38% for the collateral ligaments. These percentages could be understood as a criterion for deformation failure.

As recruitment of cruciate or collateral ligament fibres occurs according to their direction, damage localisation could be explained, at least in part, by the orientation of the ligament insertion. Further study using various orientations of the structure should be performed to evaluate the influence of the ligament insertion on failure mode.

Experimentally, ligaments failed by peeling off their insertion, by pulling out of the thickness of the cortical bone or by avulsion of the bone-ligament interface membrane (case of the collateral ligaments). These failure patterns were reproducible with the four type of ligaments tested and demonstrated the influence of bone insertion structure and of the bone-ligament hydroxyapatite gradient. This gradient could account for the brittle or ductile failure types observed for cruciate and collateral ligaments respectively. Peeling failure under traction force is interpreted as the decohesion of fibres which are linked laterally and longitudinally. This failure could then be understood as a combination of mode I and mode II failure. For bone avulsion in the cortical thickness or bone avulsion of the ligament insertion, failure was interpreted as mode I failure. We are currently conducting a histological study of the samples in order to verify these assumptions.

As all tensile tests were performed with the same velocity and as initial forces which reduced viscous effects were applied, it was difficult to conclude on the effect of strain rate in this work. Nevertheless it would be reasonable to assume that the structure was subjected to dynamic effects. The celerity calculation for the structure showed that time propagation of a traction wave was the same as the time used for the tensile test. This point should be checked in a further study with a different velocity. As we only had four post mortem anatomic subjects, we decided to apply the same load to all ligaments (with little variation in the initial force).

We were able to obtain failure data on 20 ligaments, but as all were taken from four old donors, a quantitative analysis of the results could only be representative of an elderly age group but not of the population in general. These qualitative results on the failure process occurring in ligaments under dynamic solicitations did however allow validation of our experimental device. Further tests using a more representative sample will be needed to determine the failure behaviour properties of ligaments.

CONCLUSION

A new experimental device to perform tensile traction tests on soft tissues was designed and produced. This new device allows both static and dynamic loadings for all ligaments of the knee with various orientations of the ligaments during loading. Twenty ligaments were subjected to tensile loading in the direction of their fibre axes with a constant velocity of 2m/s. The cruciate ligaments were found to be stronger than the collateral ligaments. A criterion of deformation failure can be postulated for cruciate and collateral ligaments. For all cruciate ligaments, failure was very brittle and occurred in a two-step process at the tibial insertion. These steps, identified by peaks on the force-time curve, corresponded to failure of the two main fibre bundles of the ligament structure, with peeling failure and cortical bone avulsion. For all collateral ligaments, failure was more ductile and occurred in a one-step process, with avulsion of the bone-ligaments interface.

In addition to precise control of the loading force and its velocity delivered with a hydraulic actuator throughout the loading phase, the newly developed apparatus can fix the orientation of the ligament before testing, a feature that should be most useful for future experiments. Viscous behaviour (with or without damage) and fatigue behaviour can be studied with relaxation or harmonic solicitation. This allows following the course of damage as a function of controlled loading force.

Particularly, loading can be stopped at a characteristic step (end of the elastic phase, beginning of the damage phase, end of the first peak in the strain/stress curve...) to study the structural integrity of the ligaments. This should usefully contribute to understanding the damage phase and help integrate this mechanism into the damage behaviour law. These possibilities should be combined with a histological study of the sample. Finally, it is noteworthy that this experimental apparatus can perform experiments on a wide range of soft living tissues including muscles, tendons and other ligaments.

REFERENCES

- [1] Woo S., Mechanical properties of tendons and ligaments, *fourth international congress of biorheology*, vol. 19; pp. 285-396, 1982.
- [2] Woo S., Young E., Kwan M., Fundamental studies in knee ligament mechanics, *Knee Ligaments : Structure, Function, Injury and Repair*, Edited by D. Daniel, by Raven Press, pp. 115-134, 1990.
- [3] Woo S., Hollis M., Adams O., Lyon M., Takai S., Tensile properties of the human femur anterior cruciate ligament-tibia complex, the effects of specimen age and orientation, *The American Journal of Sport and Medicine*, vol. 29, No. 3, pp. 217-225, 1991.
- [4] Mommersteeg T.J.A., The effect of variables relative insertion orientation of human knee bone-ligament-bone complex on the tensile stiffness. *Journal of Biomechanic*, vol. 28, n°6, pp. 745-752, 1995.
- [5] Mommersteeg T.J.A., Characterisation of the mechanical behavior of human knee ligaments : a numerical experimental approach. *Journal of Biomechanic*, vol. 29, n°2, pp. 151-160, 1996.
- [6] Butler D.L, Noyes F.R, Grood E. S., Measurement of the mechanical properties of the ligaments. *CRC Handbook of Engineering in Medicine and Biology*, Section B: Instruments and Measurements, 1992;
- [7] Liao H., Belkoff S., A failure model for ligaments, *Journal of Biomechanic*, vol. 32, pp. 183-188, 1999.
- [8] Hurschler C., Loitz-Ramage B., Vanderby R., , A structurally based stress stretch relationship for tendon and ligament, *Journal of Biomechanical Engineering*, vol. 119, pp. 392-399, 1997.
- [9] Elliott D. H., The biomechanical properties of tendon in relation to muscular strength. *Annal of Physical Medicine*, vol. 9, pp. 1, 1967

- [10] Crisp J. D. C., Properties of tendons and skin. *Biomechanics, Its Foundations and objectives*, Fung, Prentice all., 1972
- [11] Viidik A., Experimental evaluation of the tensile strength of isolated rabbit tendons, *Bio. Med. Eng.*, p. 31., 1967
- [12] Viidik A., Ekholm R., , Light and electron microscopic studies of collagen fibers under strain. *Zeitung Anatomische Entwicklung Geschaft*, vol. 127, pp. 154-164, 1968
- [13] Viidik A., Functionnal properties of collagenous tissues. *International Review of Connective Tissue Research*, vol 6, Academic Press, New York, 1973.
- [14] Frisen M., Mägi M., Sonnerup L., Viidik A., Rheological analysis of soft collagenous tissue- Part 1 : Theoretical considerations. *Journal of Biomechanic*, vol. 2, pp 13-20, 1969.
- [15] Noyes F., Delucas J., Torvik P., Biomechanic of anterior cruciate ligament failure. an analysis of strain-rate sensitivity and mechanisms of failure in primates. *Journal of Bone and Joint Surgery*, Vol. 56-A, pp. 236-253, 1974.
- [16] Noyes F. R., Grood E. S., The strength of the anterior cruciate ligament in humans and rhesus monkeys. *Journal of bone and Joint Surgery*, pp.1074-1081, 1976.
- [17] Trent P., Walker P., Wolf B., Ligament length patterns, strength and rotational axes of the knee joint. *Clinical Orthopedics Related. Research*, vol. 117, pp. 263, 1976.
- [18] Trevisan D., Rhéologie dynamique des tissus biologiques. *Journal de Biophysique et de Médecine Nucleaire*, vol. 7, n° 2, pp. 75-79, 1983.
- [19] Race A., The mechanical properties of the two bundles of the human posterior cruciate ligament. *Journal of Biomechanic*, vol 27, n° 1, pp. 13-24, 1994.
- [20] Pioletti D., Viscoelastic Properties of Soft Tissues: Application to knee ligament and tendon. *Ph. D; Thesis*, Lausanne, EPFL, 1997
- [21] Diamant J., Keller A., Baer E., Litt M., Arridge R., Collagen ultrastructure and its relations to mechanical properties as a function of ageing. *Proceedings of Research Society*. London, vol. B180, pp. 293-315, 1972.
- [22] Fung Y. C., Biorheology of soft tissues. *Biorheology*, vol. 10, pp. 139-155, 1973
- [23] Dorlot J., Ait Ba Sidi M., Tremblay G., Drouin G., Load elongation behaviour of the canine anterior cruciate ligament. *Journal of Biomedical Engineering*, vol. 102, pp 190-193, 1980.
- [24] Kennedy J., Hawkins R., Willis Danylchuk K., Tensions studies of human knee ligaments. *Journal of Bone Joint Surgery*, vol. 58-A, pp. 423-436, 1976;
- [25] Haut R. C., Age-dependent influence of strain rate on the tensile failure of rat-tail tendon. *Journal of Biomechanical Engineering*, vol. 105, pp 296-299; 1983.
- [26] Danto M., Woo S. L. Y., The mechanical properties of skeletally mature rabbit anterior cruciate ligament and patellar tendon over a range of strain rate. *Journal of Orthopaedic Research*, vol 11, pp. 58-67, 1993.
- [27] Holden J., Grood E., Korvich D., Cummings J., Butler D., Bylski-Austrow D., In vivo forces in the anterior cruciate ligament : direct mesurements during walking and trotting in a quadruped. *Journal of Biomechanic*, vol. 27 (5), pp. 517-526, 1994.
- [28] Maes M., Vanhuysse V., Decraemer W., Raman E., A thermodynamically consistent constitutive equation for the elastic force-length relation of soft biological materials. *Journal of Biomechanic*, vol. 22, No. 11/12, pp. 1203-1208, 1989.
- [29] Lam T., Thomas C., Shrive N., Frank C., Sabiston C., The effects of temperature on the viscoelastic properties of the rabbit medial collateral ligament. *Journal of Biomechanical Engineering*, Vol 112, pp. 147, May 1990.
- [30] Chimich D., Shrive N., Frank C., Marchuk L., Bray R., Water content alter viscoelastic behaviour of the normal adolescent rabbit medial collateral ligament. *Journal of Biomechanic*, vol. 25 (8), pp. 831-837, 1992.
- [31] Sasaki N., Odajima S., Elongation mechanism of collagen fibrils and force strain relations of tendon at each level of structural hierarchy. *Journal of Biomechanic*, Vol. 29, No 9, pp 1131-1136, 1996.

- [32] Woo S., Mathematical modeling of ligaments and tendons, *ASME*, vol. 115, pp. 468-473, 1993.
- [33] Fung Y. C., *Biomechanics, Mechanical properties of living tissues*. Second edition, Springer, 1993.
- [34] Winckler G., *Manuel d'Anatomie Topographique et Fonctionnelle*, 2^{ème} ed. Masson, Paris, 1974.
- [35] Cesari D., Cavallero C., Effects of crash conditions on pedestrian leg kinematics & injuries based on cadaver & dummy tests., 29 proceedings of the American Association for Automotive Medicine, 1985.
- [36] Kajzer J., Cavallero C., Bonnoit J., Morjane A., Ghanouchi S., Response of the knee joint in lateral impact: Effect of bending moment. Proc. IRCOBI, pp. 105-116, 1993
- [37] Kajzer J., Cavallero C., Bonnoit J., Morjane A., Ghanouchi S., Response of the knee joint in lateral impact: Effect of shearing loads. Proc. IRCOBY, pp. 293-304, 1990
- [38] Rauch G., Allzeit B., Gotzen L., Tensile strength of the anterior cruciate ligament in dependence of age. In biomechanics of human knee ligaments, *Proceedings of the European society of Biomechanics*, University of Ulm, Germany, pp. 24, 1987
- [39] Wasmer G., Hagen F. W., Mittlemeier Th., Bergmann M., Hofman G. O., In biomechanics of human knee ligaments, *Proc. European society of Biomechanics*, University of Ulm, Germany, pp. 25, 1987.
- [40] Prietto M. P., Bain J. R., Stonebrook S. N. Settleage R. A., Tensile strength of the human posterior cruciate ligament (PCL). *Trans Orthopaedic Research Society*, vol. 17, pp. 124, 1992.
- [41] Piekarski, K., Fracture of bone. *Journal of Applied Physics*, 41, 1, 215-223, 1970.
- [42] Kajzer J., Schroeder G., Ishikawa H., Matsui Y., Bosch U., Shearing and bending effects at the knee joint at high speed lateral loading. *STAPP car crash conference*, n°973326, pp. 151-165, 1997.
- [43] Huiskes R., Kremers J., Lange A., Woltring H. J., Selvik G., Van Rens J. G., Analytical stereophotogrammetric determination of three dimensional knee joint geometry. *Journal of Biomechanic* vol. 18, N. 8, pp. 559-570, 1985.