ALIGNMENT OF LOWRE EXTREMITIES. TORSIONAL ABNORMALITIES. CLASSIFICATION .

ALIGNMENT OF LOWRE EXTREMITIES.

Lower extremities of human beings, like the rest of their anatomy, are subject to profound changes from their foetal period until they reach the adult age, and these changes not only affect their size but also their morphology, proportion and special orientation.

Racial and inheritance factors play a primal role in predefining what the final morphology of lower extremity will be, although it might receive influences from an enormous quantity of endogenous and exogenous factors, which could also even modify, within certain limits, the definitive theoretical result.

This multiplicity of congenital or acquired factors justify the appearance of a huge number of structural variants at the end of the individual's growing period, which derive from the alignment combinations of the different anatomical segments of the three spaces plane. This set of alignments defines the shape of the limb and, therefore, its biomechanical behaviour towards different physical demands during its function.

Hence, when dealing with the influence of biomechanical aspects in the etiopathogenesis of all lower extremities conditions, from a statistical viewpoint, it is esteemed absolutely necessary to gather beforehand the different subjects submitted to the study into certain predefined groups, in order to randomize them properly. These predefined groups are going to be called morphotypes. When revising the existing literature regarding the possible participation of lower extremity alignment as a mechanical factor tending to develop a certain pathology, we are seized by the fact that whereas we find abundant information on the influence of misalignments on the frontal plane and some fewer on the saggital plane, publications related to the influences of torsional alignment are very scarce, especially when referring to knee disorders.

► LOWER EXTREMITY EVOLUTION

Depending on alignment changes on the lower extremity that take place during fetal life, the articulation of the hip shows in the newborn an anteversion of the femoral neck with a mean value of 45^0 as well as a valguism of 150^0 The behavior of the knee is a varus lower than 15^0 while the tibia show a coronal position oscillating between 20^0 of internal rotation and 3^0 of external tibia torsion. The foot provides for internal rotation behaviors with adducted forefoot, showing a submalleolar detorsion angle around 14^0 .

The ulterior development of lower extremity will produce a progressive decrease of femoral neck anteversion and valguism reaching at the age of 12-14 their definitive values, placed between 10 and 15^0 and 120 - 130^0 respectively .

In the knee, the initial varus will usually disappear around 6 and 24 months of life.

From that moment a physiological valgus stage is engaged, with maximum values of valgus reached around 4 years old, which progress towards normal alignment that in habitual conditions will be reached between the age of 7 and 23.

Internal tibia torsion, in case it exists, will go under a process of detorsion in the opposite sense of that experimented by the femoral neck, in other words, in an external direction.

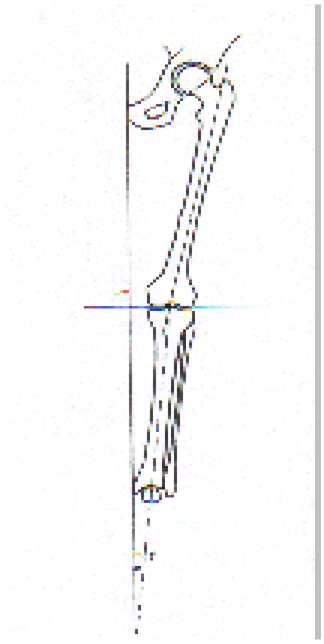
This detorsion process takes place mainly in the supratuberositary area of the upper tibia epiphysis, and all internal tibia torsion, which persists after the age of 3-4, will have to be construed as a pathology.The behavior of the foot in internal rotation and adducted forefoot will disappear progressively during the first months of life, whereas the talus declination angle will present a progressive reduction of its value. The submalleolar detorsion angle will decrease from its initial value of 14^0 until reaching an average of 3^0 in adults.

Independently from constitution or/and racial factors that modulate the repercussion of different elements **(** articular capsule, muscular action, biped standing, etc**.)**, in the evolution of different axis, other causes such as the person's weight, corporal attitude and practice of certain sports will also influence the consecution of definitive alignment of the extremity.

* FRONTAL PLANE

The alignment of the frontal plane is perfectly defined both by its magnitude and by the repercussion of its deviations in varus or in valgus in the degenerative pathology or for surcharge of the knee joint.

There are several available methods for their measurement, based on the anatomical axis of the limb or the angles that shape the axis of the shaft of femur and tibia, with an average value of 7^0 or in the assessment regarding the mechanical axis of the limb represented by a line joining the center of the femoral neck with the knee center and the ankle with the knee center, not withstanding that in Physiological conditions the mechanical axis forms an angle with the vertical of 3^0 **.** ▼

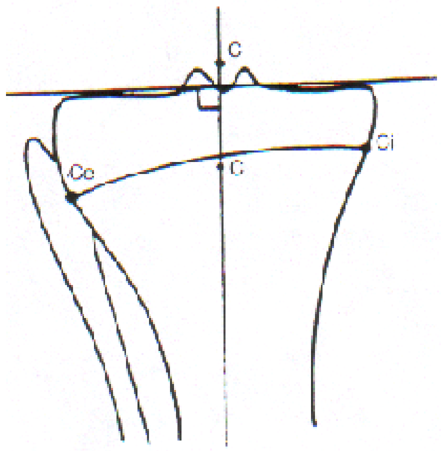


(Telemetry Image of Lower Extremity to Measure in the Anteroposterior Plane)

Furthermore, we must trace over the tibia the mechanical tibia angle between the mechanical axis and the tangent line of the tibial plateau, which in normal conditions must be of 90^0 the deviation of this angle will determine the constitutional tibia varus or valgus.

Due to the wear or usury of internal tibia plateau in cases of prolonged genus varus it is difficult Sometimes to find this mechanical tibia angle, as the tangents do not coincide with both tibia plateaus.

Thus, we recommend using the proximal epiphyseal axis, defined as the line that goes through both tibia spines and the conjunction center, as it is shown in this figure ▼



(Scheme of the Upper Third of the Tibia to Measure the Epiphyseiometaphysary)

* LATERAL PLANE

Measurement in the lateral plane has to be carried out in static biped standing through the calculus of the axis angle of the shaft of femur and tibia, after requiring the patient to extend as much as possible his knees. Their maximal normality value should not exceed from 5^0 .

The possibility of hyperextension of the knee or genu recorvatum is an anatomical abnormality, generally associated with a hyperlaxity usually accompanied by a patella alta.

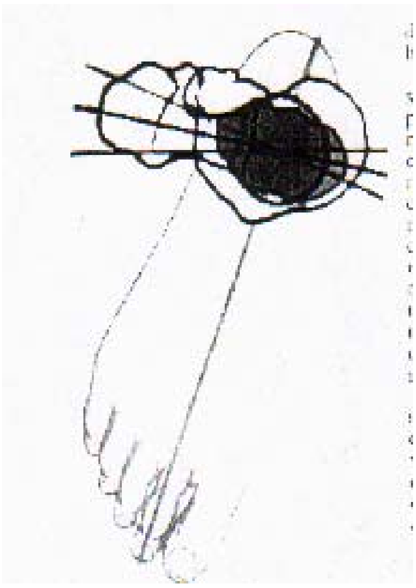
For the purpose of assessing patella alta different measurement indexes can be used. The Insall-Salvati index, which is often used and likes the rest, is not free from possible errors regarding the quadriceps state of contraction or relaxation, is obtained from the calculus of the quotient dividing the patellar tendon longitude by the maximal diameter of the patella, which in normal conditions ranges between 1.02 and 1.07.

The extension defect of the knee is usually associated in the degenerative pathology of the knee or in posttraumatic sequels, although in adolescents it might be associated with a notable shortening of the ischiotibial musculature.

All values exceeding 5^0 will imply a relevant surcharge of the femoropatellar joint.

* CORONAL PLANE

The measurement of different alignments contributing to the torsional morphotype of the limb shows greater difficulties than the previous described planes▼



(Schematic Representation of Coronal Alignment of Lower Extremity)

Despite **Netter's** maneuver in the child's hip provides an idea of the magnitude of the femoral neck anteversion with an error margin of **± 10%,** its applicability in obese individuals or suffering from a degenerative disease in the hip lacks reliability.

Netter's maneuver is carried out with the subject in a prone stance on the exmination table with the knee bended 90^0 while one of the examiner's hand, taking hold of the lower extremity leg to be explored, is placing an internal rotation of the hip, the other hand is leaning on the trochanter zone to note in which moment of the rotation the maximum lateralization is perceived it is to say, prominence of the major trochanter. The angle formed at this moment by the leg regarding the examining table plane will be the value of femoral neck anteversion.

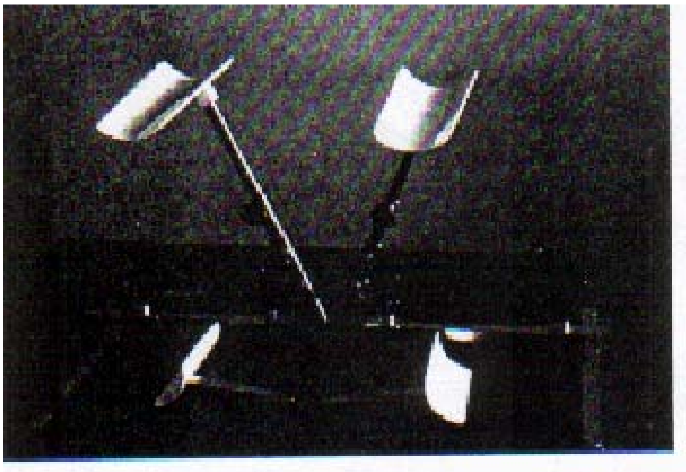
In the tibia, exploration of the patient in a sitting position, with his legs dangling off the edge of the examining table, allows us to assess quite exactly the axis formed by a bimalleolar line and the frontal axis of the knee, which will approximately represent the value of tibial torsion.

When attempting to quantify exactly the angular values of the lower extremity we are faced with a difficult problem to solve.

In radiological exploration, the projection is a three-dimensional structure plane offering great possibilities for erroneous measurement regarding the extremity position.

Radiological technique must be very accurate as for the position of the extremity during the shooting and, on the other hand, direct measurement on the simple radiography is only valid in anteroposterior and lateral axis of the knee.

Dunn's positioning device to calculate femoral anteversion.



(Dunn’s Positioning Device to Calculate Femoral Anteversion)

For a simple radiological measure of coronal values we must resort to an instrumented measurement, always starting from a rigorous radiological technique.

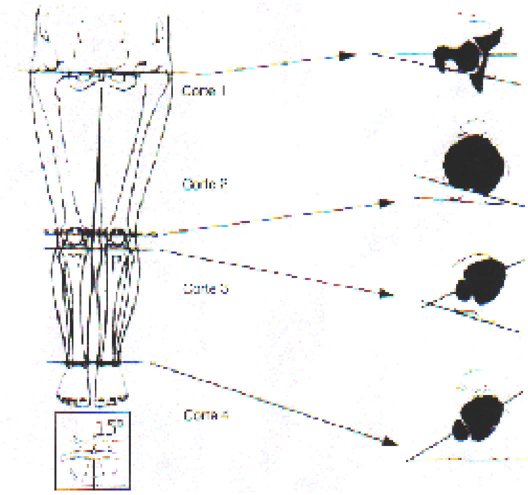
In the hip, when using a Dunn's or Dunlap's positioning device an anteroposterior and axial projection for the hip must be performed. Projection values obtained from the measurement of the Cervicodiaphysiary angles in both planes will be translated into a conversion table that will provide a real value of valgus and anteversion of femoral neck.

In the leg, a lower extremity radiograph with the subject sitting, knee bended and the foot leaned on the radiological screen, will provide for a direct value of the tibial torsion, if we have previously used metal signalers to mark the limb, as indicated by **Fernández Sabater**.

Both techniques, with a mild margin of error if carried out properly, prove to be of difficult universal application for their complexity, laboriousness and need of harshness in their execution.

Thanks to the computed tomography (CT), having the chance to perform different sectionings along the limb of the subject without additional instrumentation, we have been able to interpret much more clearly the reality of the shape of the limb in the coronal plane.

Torsional scanner, non-exempt from measurement risks, allows the performance of Sectionings at an acetabulum level **(acetabulum anteversion),** femoral neck and femoral condyles **(fermoral anteversion),** supratuberositary tibial tableau and malleolar zone **(tibial torsion)** and malleolar zone and foot **(submalleolar detorsion),** independently from other knee measurements regarding the coronal position of its extensor apparatus▼ .



(Schematic of the Measurement of Femoral Anteversion and Tibial Torsion . Section 1, Section 2, Section 3, Section4)

Having data from the anteversion of femoral neck and tibial torsion allows obtaining the tibiofemoral index, though the subtraction of both values **(ETT less AVF)** and that if having an approximate mean value of 20^0 will be really useful in order to classify torsional groups.

► TORSIONAL MORPHOTYPE:

The possibility of interrelation torsional abnormalities with degenerative pathology or by surcharge of lower extremities compels a previous classification of subjects, for the purpose of favoring an appropriate randomization of different series.

Knowing the difficulty beforehand and being aware of the defects involving simplicity of a schematic classification in a subject engaging so many variables, we have suggest and used for a long time the following classification groups .

► CLASSIFICATION OF PATIENTS

The patients, object of study whose alignment abnormality of the limb is being correlated with a determinate pathology, or in planning a surgical treatment of misalignment correction, must be object of a protocol study, first clinically and later on throughout imaging according to a systematic constant allowing us the proper classification.

Therefore, we start with clinical exploration of the naked patient in static biped standing position, followed by a barefoot ambulation over a plain ground.

Afterwards, on the examining stretcher where we evaluate the joint balance of the hip, knee and foot.

The following tests are required**:** **1)** Telemetric radiography of lower extremities in anteroposterior projection in biped standing position, with patellae aligned to the zenith, maximal extension of knee profile and axial extension of the patella at 15^0 of flexion.

2) Torsion CT of lower extremities with sectioning in the femoral neck, femoral condyles, tibial tableau, maleolla and foot.

Additional radiographies or CT sectionings required regarding could complete the examination each specific case .

Measurement results obtained must be adequately recorded in their corresponding protocol page.

► (Protocol page for the record of angular values and torsional groups)

Valgus ⟸ **Normal Axis** ⟹ **Varus**

* Anatomical axis ☐ **>10^0** ☐ **5-10^0** ☐ **1-5^0** ☐ **1-7^0** ☐ **1-5^0** ☐ **5-10^0** ☐ **>10^0** **Q angle**

* Mechanical axis ☐ **>10^0** ☐ **5-10^0** ☐ **1-5^0** ☐ **1-7^0** ☐ **1-5^0** ☐ **5-10^0** ☐ **>10^0**

* Tibial slope ☐ **< 3^0** ☐ **3-5^0** ☐ **5-7^0** ☐ **>7^0** **Tibial proximal metaepphysiary**

***Arthrosical signs. Ahlback**	**Torsional group**	**AVF**	**TT**	**Femoral anteversion**
1) Interline clamp	**1**	**N**	**N**	**Femoral torsion**
2) Interline absence	**2**	**15^0-30^0**	**35^0-50^0**	**Tibial torsion**
3) Minor abrasion of subchondral bone	**3**	**N**	**>35^0**	**Femorotibial index**
4) Moderate abrasion (0.5 to 1 cm)	**4**	**15^0-30^0**	**>50^0**	
5) Severe abrasion, often subluxation	**5**	**N 0 < 10^0**	**< 25^0**	**TA-GT distance**

* **Patellar alignment** ☐ **Normocentered** ☐ **Minimal subluxation** ☐ **Moderate subluxation** ☐ **Maximal subdislocation**

* **Femoropatellar arthrosis** ☐ Yes ☐ No Wiberg ☐ 1 ☐ 2 ☐ 3

* **Intercondylar sulcus** ☐ Flat ☐ Profound ☐ Normal

DETERMINATION OF CONTACT AREAS.

DETERMINATION OF CONTACT AREAS.

The Study of the Biomechanics of the Human Patello-Femoral Joint is necessary to understand or solve clinical problems such as chondromalacia, the lateral patellar compression syndrome, and recurrent subluxation or dislocation.

To address these problems, **Many Experimental and Analytical Studies have been Performed.** In the last twenty years has focused on determining contact areas and stresses in joint, using experimental methods and theoretical analyses.

(1) Measurement methods for contact surface or load in a normal knee.

(A) EXPERIMENTAL METHODS: -

1) Radiographic and Sectioning Techniques : .

The earliest studies of diarthodial joint contact were performed radiographically, where "points" of contact was determined. Radiographic techniques have alsobeen employed in conjunction with radio-opaque solutions injected into the joint (e.g., **Kettekamp and Jacobs**, **1972**[165] **.** Used this method in a survey on 14 cadaveric knees where he injected a radiopaque contrast and applied some compression force of around 8 kg. Load areas in extension were approximately 4.68 cm^2 in the internal plateau and 2.97 in the external plateau. **Maquet 1975**[166] . Sectioning Techniques have been used to determine joint contact areas. used the same method but he applied stronger forces to the knee, ranging between 225 and 250 Kg. He found contact surfaces in the internal plateau of 20.13 cm^2.

The same author believes that differences from Kettelkamp's Researches are due to the fact that a load increase produces an increase of the load area so that stresses on articular surfaces are not modified. Another radiographic technique is stereophotogammetry.

Wiberg sectioned frozen knee joint to analyze contact in the patellofemoral joint [(167)] **.** While this technique can be used for only one joint position; the results reported by Wiberg were in agreement with subsequent studies uses other methods.

2) Dye staining techniques:

Greenwald and O' Connor 1971 determined contact areas in the hip joint a dye staining method [(168)] **.** In this method, the cartilage surfaces are first allowed to absorb a chemical agent prior to applying a load on the joint. While the joint is loaded, it is exposed to a second liquid chemical agent, which reacts with the first agent in cartilage to produce a colored stain in all those regions of the articular surfaces not in contact. Subsequently, the joint is washed with normal saline and the surfaces are removed to allow viewing of the staining. A similar approach was employed by **Goodfellow et al.1976** [(169)] for studies of the knee joint and **Moran et al 1985** for studies of the finger joint [(170)] . **Matthews et al 1977** applied methylene blue on the retropatellar surface prior to loading the joint, and then assessed the size and location of the imprint of the patella on the opposing trochlear surface [(171)] **.**

3) Casting techniques:

In this method, a casting material is injected into the joint while still a liquid, either prior or subsequent to load application. The most common casting materials used are methylmethacrylate and silicone rubber. Probably the earliest casting study, was performed by **Walker** and **Hajeck 1972** on the tibiofemoral joint [172] **.** Recently, **Yao** and **Seedhom 1991** introduced the 3S technique in which a carbon and silicon oil powders is introduced. where a silicon oil-carbon black powder suspension is squeezed out the regions of contact between the joint surfaces following loading [173] **.**

4) Pressure methods :

When calculation contact surfaces some authors used systems of sensitive film,"Prescale" type that will be analyzed thoroughly in the section of loads and stresses calculation.

(B) ANALYTICAL METHODS:-

Surface Proximity Techniques:

Scherrer et al.1979 introduced a joint contact determination method based on the calculation of the relative proximity of the articular surface at various joint positions [174] **.** A similar analysis was performed by **Soslowsky et al 1992** [175] using close- range stereophotogrammetry **(SPG)** for surface Topography [176,1177] and kinematics Measurements. Most recently, **Kwak et al 1993** reported on the contact areas and cartilage thickness distribution in greyhound patella, distal femurs and tibial plateaus using SPG [178] **. Figure (a)** demonstrates the relative position of the patella and distal femoral articular surface at 60 and 120 degrees of felxion, while **Figure (b)** displays the contact areas on the patella at 30, 60 and 90 degrees.

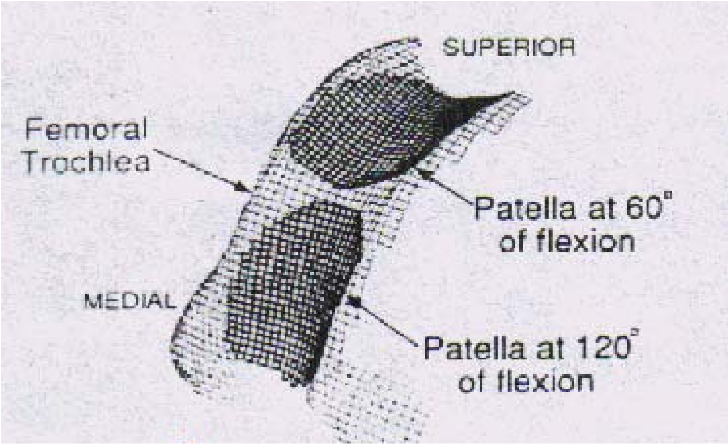


Fig. (a)

Typical configuration of the relative position of the greyhound Patella and distal femoral articular surface at 60 and 120 degree of flexion, as obtained with stereophotogrammetry.

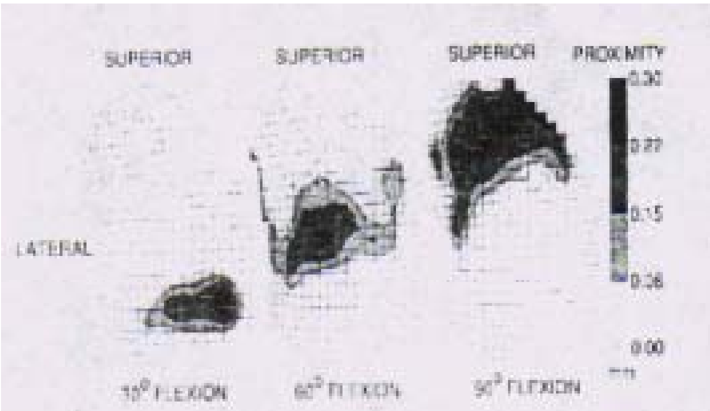


Fig. (b)

Patella contacts areas as proximity maps, 30, 60, and 90 degrees of flexion.

(2) Measurement Methods for Stress in Articular Surfaces

(A) EXPERIMENTAL METHODS: -

Pressure Measurement Techniques.

In addition to measuring contact areas, investigators have also been interested in measuring contact stresses or pressures. **Ferguson et al 1979** measured contact stresses in the patellofemoral joint using miniature piezoresistive contact pressure transducers implanted at various locations on the retropatellar cartilage [(179)]. This type of transducer can be used for measuring transient stress responses at discrete locations in the joint, providing a history of the time-dependent response of cartilage, which is known to behave in a viscoelastic manner. Piezoresistive transducers were also used by **Brown** and **Shaw 1983** to look at contact stresses in the hip[(180)] and on the femoral condyle surfaces[(181)] **.**

Fukubayashi and **Kurosawa**[(182)] used pressure sensitive film **(**by Fuji Film Co., Ltd., Tokyo.**)** to measure contact pressures in the joint directly, by inserting the film between the contacting articular surfaces. They studied pressures when applying 1000 and 1500 N loads in cadaveric knees before and after a meniscetomy. This pressure sensitive film consists of two polyester sheets which, when pressure against each other, will produce a red stain whose intensity depends on the applied pressure. As the film is sensitive to humidity, and the film is approximately 0.2 to 0.3 millimeters thick, the Fuji pressure-sensitive film has been used extensively. **Huberti** and **Hayes 1984** used pressure sensitive film to analyze the influence of Q-angle and tendofemoral Contact[(183)] as well as effects of capsular reconstructive procedures[184] on patellofemoral contact pressure. Similarly, **Brown** et al **..** studied contact stress aberration in the presence of osteochondral defects **1991**[185]**, Haut** measured patellofemoral contact pressures during impact loading of the knee[186] . A similar technique was employed by **Ahmed 1983** who studied pressure distributions on the tibia[187] and the retrepatellar surface[188] **.**

Hirokawa and Matsumura[189] made measurements of patello-femoral joint contact pressure using a specially designed film that senses pressure by the variation of electrostatic capacity under conditions simulating successive variations of the knee flexion angle and thigh muscle forces for three activities: level walking and climbing and descending stairs. **Koh et al**[190] measured and described the dynamic, in vivo, three dimensional Tracking pattern of the patella for one normal male subject. In their experiment, intracortical pins were inserted into the patella, tibia and femur, and the markers triad were attached to the pins, respectively, the subject performed seating squatting knee flexion/extension, and maximum voluntary quadriceps contractions. **The results were found to be qualitatively in agreement with those of an in vitro study by Van Kampen and Huiskes**[191] **, verifying the data collected from cadaveric specimens.**

Recently a comparison study of the above techniques has been completed (dye staining, silicone rubber casting. Fuji pressure-sensitive film, SPG, **1993**)[192] **.** It was found that dye staining could over- estimate contact areas while silicone rubber casting tended to underestimate contact areas, particularly in high congruent articulations, Fuji film and SPG provided very consistent results.

B) Theoretical analyses :

Theses methods are based on the analysis of contact and stress surfaces starting from the theoretical models to which a series of formulas and mathematical and physical properties are applied.

In available literature we find that **Hirsch** used in **1944** the Hertz contact theory for contacting elastic spheres to model cartilage indentation[193] **.** **Askew** and **Mow 1978** analyzed the problem of a stationary parabolically-distributed normal surface traction acting on a layered transversely isotropic elastic medium to assess the function of the stiff surface layer of cartilage[194] **.** More recently, **Eberhardt et al 1990, 1991**[195,196] **,** developed a solution for the contact problem of normal and tangential loading of elastic spheres, with either one or two isotropic elastic layers to model cartilage. Using the biphasic model for cartilage[197] **,** **Mow** and **Lai 1980** calculated stresses in a cartilage layer subjected to a moving parabolically-distributed normal surface traction[198] **.** **Armastrong 1986** studied the contact problem between a cylinder and an elastic layer resting on a rigid foundation, using a thin layer asymptotic analysis for compressible and incompressible material[199] **.** **Ateshian et al 1992-1993** performed an asymptotic analysis of the contact of two spherical isotropic biphasic layer subjected to a step normal load subsequent to the depletion of the thin lubricant film between the surface[200,201] **.** Other studies have made use of joint mathematical models, which incorporate realistic, geometric data, to predict contact areas and stresses in joint by **Hirokawa 1991**[202] designed a three-dimensional mathematical model of the patello-femoral joint that takes into account the geometries of the joint and the mechanical properties of the patellar ligament. Although the basic assumptions and simplification in the process of this modelling were referred in the tibio-femoral model of **Wismans et al**[203]**.** **Hirokawa**[204] performed a two-dimensional mathematical model analysis in a Horizontal plane to investigate the influence of both insertion and strength/elasticity of each extensor **(** muscle, tendon, and fascia **)** in patello-femoral disorders, as well as to investigate the effect of operative procedures on the disorders, in combination with experimental design theory for analyzing mutually correlated influences.

Finite Elements Models

Finite element models consist on the generation of theoretical articular surfaces through computer processes and subsequent analysis. Theses models are used to calculate contact and A stress surfaces and allows assessment of the influence that changes on the articular model **(** Such as torsion angles, surface thickness, etc.**)** exert on different stress and contact surfaces values.

Many authors have studied and analyzed the role of menisci and the stress they suffer.Works by **Sauren**[205] **(1984), Hefty (1987)** are important because they designed symmetrical toroidal models to study low frictions of the meniscus [206] **.**

Three-dimensional models including contact areas in the study have been developed **by Aspen (1985)** to analyze menisci [207] **.** Nowadays different filed are being developed in the study of articulations with finite elements: **1).** Analysis of finite elements in 3-D and 2D. **2).** Automated methods for 3-D models control. **3).** Integration of SPG (stereophotogammetry) in finite elements systems in 3-D and finally, the resolution of algorithms for time-dependant problems within the viscoelastic features by means of supercomputation methods. All theses methods have been developed to improve knowledge of articular surfaces and tribology.

Materials.

Materials.

A) Specimens.

B) Simulator.

1- Gait Phases.
2- Forces to be Simulated.

* Muscular Forces :-

.... Extensor Force:Quadriceps.
.... Flexor Force: Femoral Biceps.

3- Parameters Controlled by the Simulator.

C) Strain Gauge.

D) A) SPECIMENS.

Six fresh, frozen human knee joints from six individuals were used. The specimens had no previous surgery, and had macroscopically intact cartilage, radiographically normal bone structure, and intact joint capsule. The specimens were carefully dissected, leaving only the femur, tibia, fibula, and joint capsule, and Quadriceps tendon, were quick-frozen to **– 40^0 C**. Immediately prior to testing, the specimens were thawed to room temperature.

B) SIMULATOR.

Simulator Type II. Forces External Applied.

Reproduce Gait.

1) Gait Phases:

Before beginning to analyze the forces to be simulated it is important to know which phases divide the gait cycle in order to be able to analyze afterwards what happens in each force in the different stages.

Human Gait Goes Along Three Phases:-

1) Single-Limb Support Phase.
2) Swing Phase.
3) Double-Limb Support Phase.

1) Single-Limb Loading Phase:

It begins when the heel of the front limb touches the ground and it finishes when foot toes release contact with it. The loading phase takes approximately 60% of the cycle and is divided into **5 Periods:**

I. **Heel Strike or Support:** The loading phase begins when the heel of the front limb touches the ground. This period constitutes 15% of the gait cycle.

II. **Foot Flat:** When after heel collision, the sole of the foot contacts the ground and the limb holds all body weight; it takes the following 15% of gait.

III. **Midstance Limb Loading:** Once the foot is firmly fixed on the ground, the leg and the rest of the body initiates a movement ahead, through hip and knee flexion, so that the centre of gravity of the body is located directly over theknee joint or supporting area.

IV. **Heel-Off:** This phase begins when the heel starts to leave the ground, keeping contact with the frontal part of the foot and toes. This period, also called ascendant drive, is initiated by lifting or separating the heel and constitutes the following 25% of the gait cycle.

V. **Toe-Off:** Immediately after the heel leaves the ground, the body is driven towards by a powerful contraction of the calf muscles. This phase finishes when the toes finally loose all contact with the ground and the limb begins to start the swing phase. The length of this acceleration phase is a 5% of the gait cycle and its end indicates that a 60% of the cycle has been completed.

2) Swing Phase:

It starts when foot toes leave the ground and ends when the heel touches it again. The swing phase takes 40% of the cycle. **It is divided into:-**

A) Initial Oscillation or Acceleration: the swing phase begins when foot toes leave-off the ground. At this point, limb movement has to speed up to be able to leave the ground and place ahead of the body to get ready for the next heel hold. This period takes a 10% of the oscillation phase.

B) Average Swing: It takes place when the limb goes directly ahead of the body. At this point the limb must withdraw sufficiently to avoid contact between the other foot and the ground. The period takes the 80% of the swing phase.

C) Deceleration / Slow Down: It follows the average swing phase, when the limb's ahead movement is restrained in order to control foot position right after the support phase. This period takes the final 10% of the oscillation phase.

3) Double Support Phase:

During normal gait there is a double support period, during which both limbs keep contact with the ground at the same time. This happens between the forefoot leave-off of a limb and the periods of heel strike, and foot flat support in the contralateral limb. The length of double support is directly related to the cadence **(Gait Rhythm),** therefore as cadence decreases, time spent in double support increases; and when gait speed increases; time in double support decreases. Absence of double support is used to distinguish between walking and running.

2) Forces To Be Simulated:

In order to be able to simulate a correct kinematics and kinetics of gait we consider a requirement to be able to simulate the following forces.

i. Ground reaction force.
ii. Mediolateral reaction forces.
iii. Muscular forces:-

- **Extensor Force :- Quadriceps.**
- **Flexor Force :- Femoral Biceps.**

Vertical Reaction Force. The following results were obtained. **(Figure 1).**

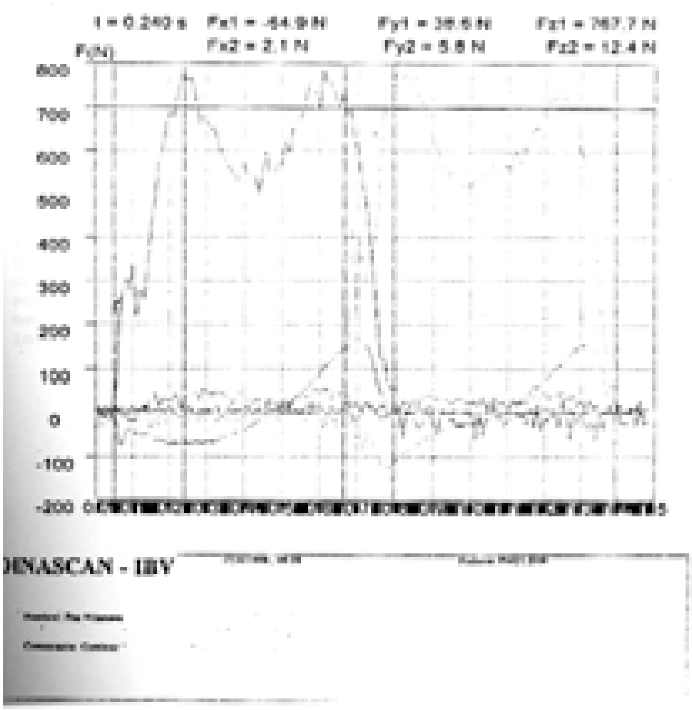


Figure 1: Graphics of reaction forces obtained with double platform of force.

- Thanks to the tests performed, we could obtain quantitative and qualitative data of different patterns of gait, taking the normoaligned knee gait patter to design the simulator.

Muscular Forces.

Muscular forces selected for simulation were the following:-

-- **Leg Extensor Group/Set: Quadriceps.**

-- **Leg Flexor Group: Femoral Biceps.**

These two muscular groups were selected because they fall within the most studied literature. We find an enormous amount of electromyographycal studies where the percentage of muscular contraction at each gait moment was calculated without giving any quantitative figures. To calculate forces used in our simulator we use Prilutski et al. data **(J. Biomechanics 29 (4), 405-415, 1996).**

For the tests performed we took obtained and verified figures within available literature although at the moment we are developing a project to calculate muscular forces through electromyography and platform of forces. Data obtained in the future thanks to this project will be compared with the literature and will be taken into account at the moment of carrying out new simulation tests. Our forces control system through hydropneumatic pistons allows reprogramming them. Thus we can change force figures at any time without altering its operation. **(Figures 2-3).**

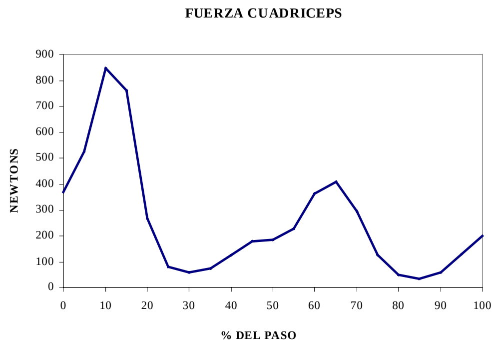


Figure 2: Graphics of force in the quadriceps depending on the gait moment.

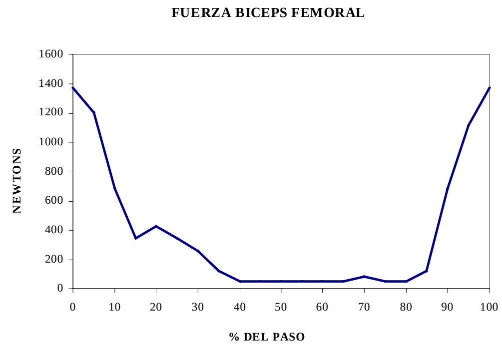


Figure 3: Graphics of force of femoral biceps depending on the gait moment.

The Simulator we wanted to design had to be able to simulate the following features:-

1. Simulating gait kinematics.
2. Reproducing quadriceps, femoral biceps forces.
3. Reproducing ground reaction forces and mediolateral forces
4. Allowing control of previous parameters and their modification if necessary.
5. Allowing modification of limb axis.

The simulator design corresponds to de so-called **Type I Simulators**, in which forces are externally applied on muscular, tendon and osseous elements.

This sort of simulators is employed to perform experimental studies with knee specimens.

The following illustrations are schematic drawings of the original simulator **(Figures 4-5).**

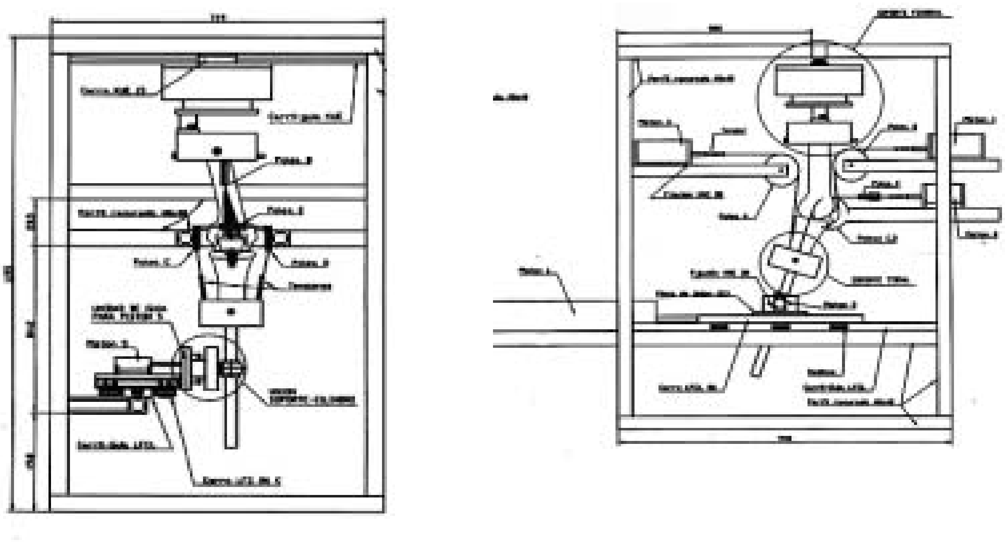


Figure 4: Schemes Of The Knee Simulator

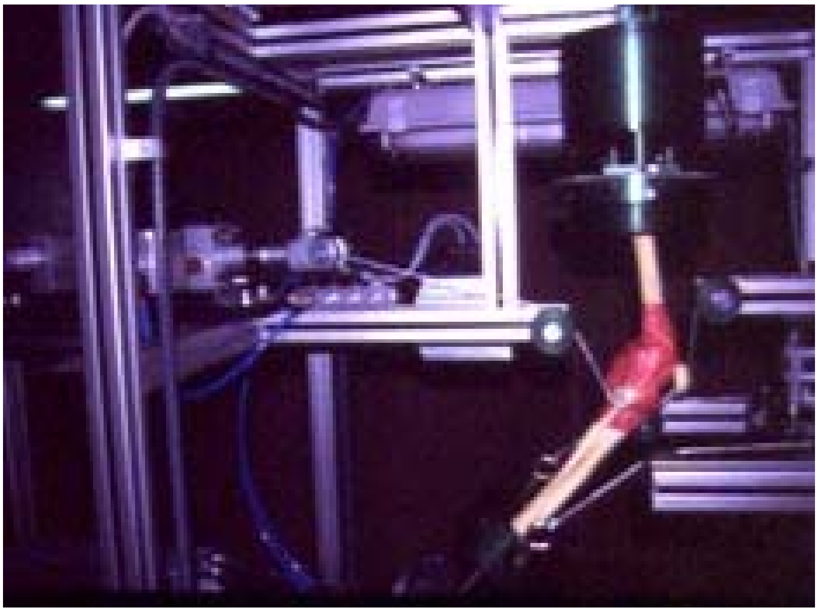


Figure 5: Picture Of The Simulator While Carrying Out A Test With Phantomas.

3. Parameters Controlled By The Simulator:

Gait Kinematics Control.

Gait kinematics control is carried out through the efforts performed by the hydropneumatic cylinder **N^0 1**. The simulator's central computer controls efforts carried out. Through the system's program we can introduce different angular curves to simulate situations such as gait, jumping, etc.

During each step, the knee performs four movement arches, with alternate flexion and extension. Usual knee movement during gait allows some flexion angles ranging between 0-70^0. The exact limits of each flexion or extension change slightly according to different tests.

These Phases could be divided into:

1) Initial Foot Contact With The Ground: During this phase the knee is flexed some 5^0. This initial angle will range on each individual between –2^0 and + 5^0. After the initial contact phase there is a quick flexion up to 18^0 of flexion corresponding to the 15% of gait. A gradual extension takes place during the rest of the contact phase until it reaches 3^0 of flexion and covers for a 40% of gait.

2) Swing Phase: At the end of the monopodal-feet phase the knee reaches some 7^0 of flexion. At the beginning of the bipodal-feet phase there is a quick flexion of the knee reaching some 63^0 and a 70%. After this quick flexion there is an equally quick extension. At the end of the swing phase there is a slow down of extension that finishes it up to 2^0 when the 97% of the cycle are completed. Afterwards the knee bends in flexion once again Reaching up to 5^0 and it enters the ground contact phase.

Maximum and minimum angles reached by the articulation vary with each individual.

In our experimental research we included gait kinematics curves accepted by international scientific literature and verified in the platform of forces of the **INEFC in Barcelona.(Figure 6).**

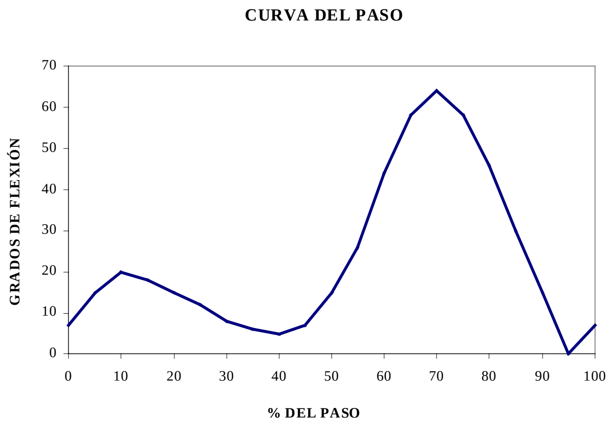


Figure 6: Scheme Of Gait Curve (Phase Curve).

(GAIT CURVE)
(FLEXION DEGREES)
(% OF GAIT)

1) Ground Reaction Forces

Ground reaction forces and mediolateral forces are controlled and reproduce by hydropneumatic cylinders **N^{0} 2 and N^{0} 5.** Figures used were the ones calculated in the **INEFC** laboratory in Barcelona by means of the platform of forces. The central computer controls value figures whose modification is possible when necessary.

(Figures 7-8).

Figure 7: Picture Of Cylinder 2 Generating Efforts That Simulate Vertical Reaction Forces.

Figure 8: Picture Of Cylinder 5 Generating Efforts That Simulate Mediolateral Reaction Forces

2) Muscular Forces.

Controlled and reproduced muscular forces are the following:

--- Extension quadriceps force. Controlled by the hydropneumatic cylinder **N^0 3.**

--- Flexion femoral biceps force. Controlled by the hydropneumatic cylinder **N^0 4.**

Value figures of efforts reproduced by the cylinders are those obtained in international literature. Monitoring is carried out through the central computer where values can be modified at any time, or where we can introduce effort curves corresponding to other situations such as jumping, running, etc. **(Figures 9-10).**

Figure 9: Picture of the Cylinder Number 3. It reproduces extension efforts of the knee produced by quadriceps muscle.

Figure 10: Picture of the Cylinder Number 3 and its connection to the phantomas.

Figure 9

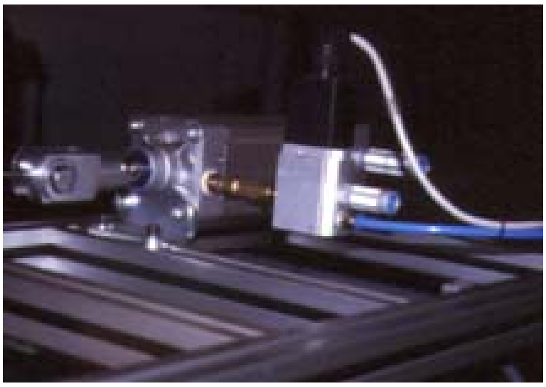


Figure 10

3) Limb Axis.

The simulator allows monitoring and modifying limb axis in the saggital and transversal planes.

- It allows monitoring:-

    ---- Varus-Valgus of the Limb.

    ---- Torsional Alignment: Femoral Anteversion and External Tibial Torsion.

- The simulators is constituted by the following elements:-

    1) **Hydropneumatic System**: The hydropneumatic system is formed by five hydropneumatic cylinders-pistons that will be in charge of producing efforts to simulate:

        a) Flexion-extension movement.

        b) Ground reaction force.

        c) Mediolateral component of ground reaction force.

        d) Extension quadriceps force.

        e) Flexion femoral biceps force.

The following section will present thoroughly each one of the elements contained in the hydropneumatic system.

A) Hydropneumatic Pistons.

Hydropneumatic pistons are the elements that allow generation of efforts through air pressure. Pistons used in our design are FESTO pistons. In order for the pistons to be able to operate properly it is necessary to have and air pressure of **8 Bar** and a **61/min** flow. Hydropneumatic pistons are fitted with a pressure-regulating valve that will sustain the required pressure at every moment at the piston input so that it exerts the necessary pressure on the cable**. (Figure11).**

Figure 11: Picture Of The Regulating Valve Of A Hydropneumatic Cylinder Of The Simulator.

The piston responsible for monitoring gait cycle must be provided with a location control system. This control system consists on an **SME-8K-LED-24** electrical proximity switch manufactured by **FESTO.** The **SME** is used to verify the location of the cylinder embolus and thus obtain the exact zero point of flexion. When the switch is off we must make a movement on the piston until reaching activation and then we will be able to carry out a system's zero.

***B) Hydropneumatic Compressor*.**

In order to obtain an **8 Bar** pressure with a **61/min** flow the simulator must be connected to an air compressor **Ingersoll-Rand ES200**. This compressor is fitted with a sound proof system that produces a maximum noise of **60Dba**. It includes a control panel with a manometer, a counter for operating time, an on-off selector switch and a light indicating if the current is connected. Furthermore it has a condensed- emptying valve and an air-service valve.

C) Efforts Transmission.

Efforts generated by pistons must be transmitted to the application spots of the forces on the specimens. In order to carry out such transmission we used a special wiring capable of enduring pressures up to **4000 N.**

Regarding the flexion system, the cable will be linked with the head of the fibula and the femoral biceps tendon. In order to avoid tiredness and distribute strain the cable is divided and joined partly to the head of the fibula and partly to the back area of the tibial plateau. **(Figur12).**

Figure 12: Picture Of The Femoral Biceps Tendon Where The Cable Connection To The Fibula Head Will Be Performed.

In the extensor muscles system, the cable will be directly joined to the quadriceps tendon through a brace. The angle between the quadriceps and the femur will be also determined by the position of a pulley. Thanks to the location of such pulley we will also obtain the **Q angle**. Cables will perform some changes in the directions at the pulleys fixed to a lateral plate, so that it is possible to make slight modifications of their position and thus correct relative positions of the muscles regarding any other bone.

D) Valve System.

Hydropneumatic pistons have a valve system to control flow and allow a regular generation of efforts.

During the entire cables cycle, both flexion and extension systems keep strained. Such strain fluctuates according to the amount of pressure allowed to the piston by the valve. Sat the same time the valve is controlled by the computer.

When strain in the piston exceeds valve tare, the latter opens and allows circulation of flow directly to the tank, and then keeps the required pressure at every moment.

The system has a non-return valve that prevents air returning to the compressor when it is being expelled from the piston due to the outer force of the cable provoked by the flexion-extension movement of the tibia. In this way all air is discharged forcedly through the pressure regulator.

2) System Of Limb Axis Modification.

The system of axis modification included in the knee simulator enables limb axis modification in**:-**

- Transversal plane: AVF, TTE.
- Saggital plane: Varus-Valgus.

Modification of the axis in the transversal plane is achieved my means of a system of Sliding rings placed on the friction areas of the femoral diaphysis. **(figure 13)**

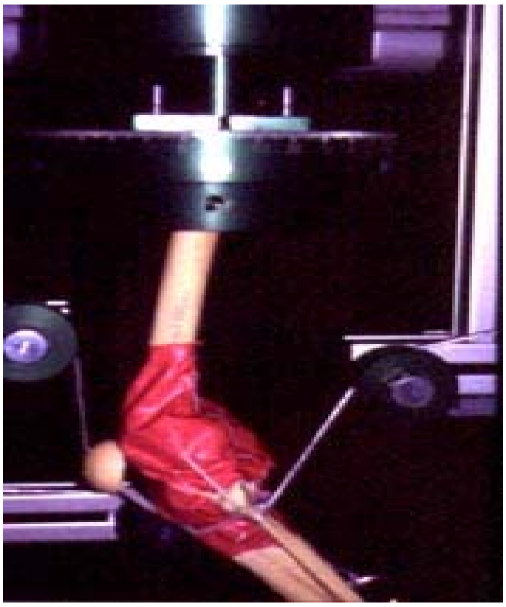


Figure 13: Picture Of The Double-Cylinder System That Enables Modification Of Femoral Anteversion.

Modification in the saggital plane is achieved by a translator-track that enables mediolateral sliding of the femur while keeping the tibia firmly fixed, so that a varus-valugs deformation is generated. **(Figure 14)** The degree of tibial torsion can be controlled with a scale located on the basis where the specimen is going to be introduced. **(Figure 15).**

Figure 14: Picture Of the System That Enables Modification Of Varus Valgus

Figure 15: Picture Of The Grading System For Tibial Torsion On The Basis Of Specimen Implantation.

3) Control System. (Figure 16).

The entire system of the simulator is controlled through a computer that enables acquisition of data inputs about knee position. After performing an analog conversion it is in charge of controlling the output and regulation valves of the pistons, which provoke the efforts of the ground reaction forces, quadriceps forces and femoral biceps forces.

A CIC-9 program, designed to transform analog inputs and to control outputs from the pistons, manages the entire control system.

Figure 16: Picture of the Simulator Control System Composed By A Control Station.

C) Strain Gauges .

Are small electrical resistors which have the property of changing their resistance under strain and which can be affixed to the surface of an object in order to measure strain in that object.

The registered deformation are transformed into differences of potential. The extensometry has been a technique used in engineering for measure of the deformation of materials, resistance's maximum, etc.the band extensiometry **(Gauge)** it constituted by a thread metallic put in a way of grilled on a material surface thermoplastic. The band extensometry to be submitted to an effort **(Either of Traction or Compression)** suffers a deformation of the thread metallic that it remains registered in the form of difference of potential. Within our unit have been used two types of strain gauge or bands extensometry.

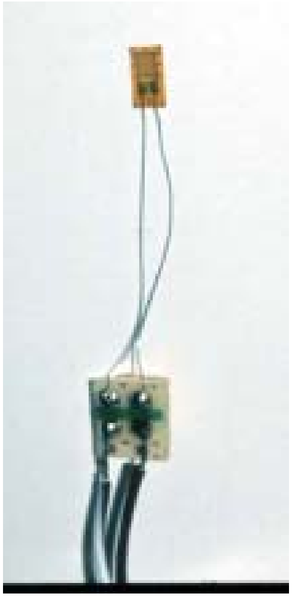
There are metal strain gauges and semi-conductors.

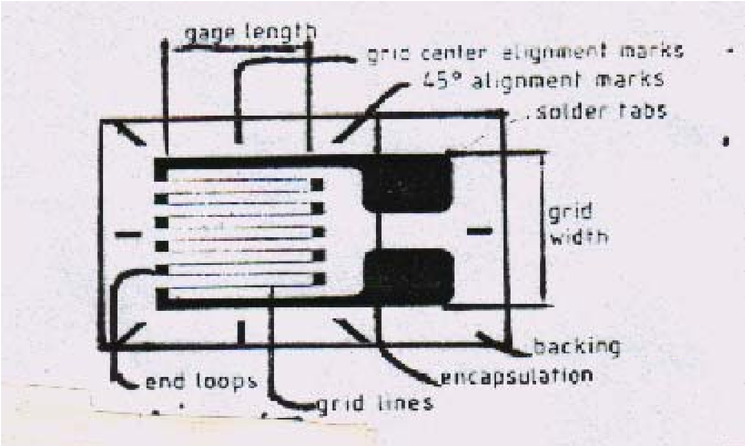
1) The Metal Ones are the larger of the two and are less sensitive to strain, and have a lower resistance (Typically 120 Ohm) However, they are easy to use, inexpensive, and the effect of temperature on the resistance is compensated by the material under testing.

2) The Semi-Conductors are smaller, more sensitive, and have a higher resistance (Typically 1000 Ohm) . These are expensive, sensitive to temperature and may be negatively affected by the environment into which they are placed .

The Terminology used for describing the strain gauge is given in this figure .

(And This of Type of the Strain Gauge that used: 2mm) .

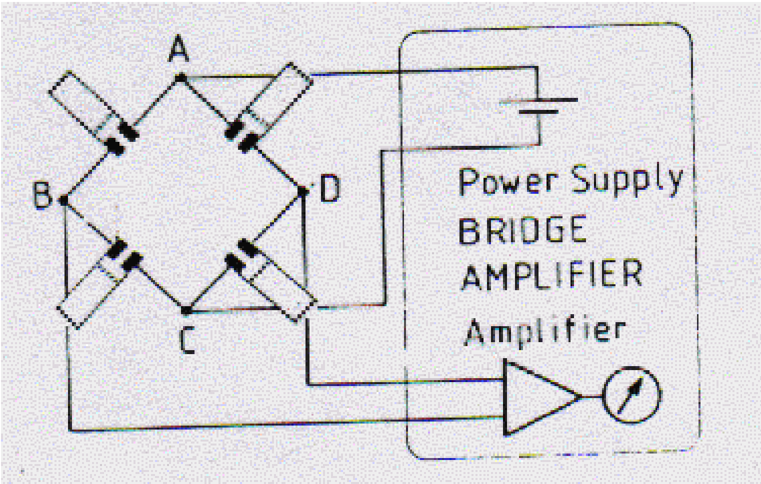




Single Strain Gauge.

Strain gauge consisting of thin wires aligned in the direction that strain will be measured, and large soldering tabs. The small marks on the strain gauges are an aid to their accurate placement on the test surface .

Strain gauge connection : The strain gauges are used exclusively to measure strain or as components in a transducer. The value measured is the increase or decrease in resistance when a part is stressed. The gauges are connected as a Wheatstone bridge as follows **:**



Wheatstone Bridge : Four strain gauges connected to form the bridge **A-B-C-D .**

A low voltage is applied between A and C . The output voltage resulting from variations in resistance in one or more of the gauge is measured between B and D after appropriate amplification .

It is possible for any or all the four strain gauges to be active, i.e. measuring strain. A Quarter bridge refers one active gauge, the other three being replaced by precision resistors.

- If two active gauge are used (A to B and B to C) and the two gauges C to D and D to A are replaced by precision resistors, we have a half bridge .
- If all four strain gauges are active, we have the conventional full bridge.

TYPICAL METALLIC FOIL STRAIN GAUGES:

A- Single gauges .

B- Rosettes of strain gauges.

C- Stacked gauges at **45^0 .**

D- Rosette for pressure transducer.

In Biomechanics, it is very easy to assess strain using strain gauges. The keywords of this : Stress, Strain, Material properties, Strain gauge, Mechanical engineering **.**

STRESS AND STRAIN :

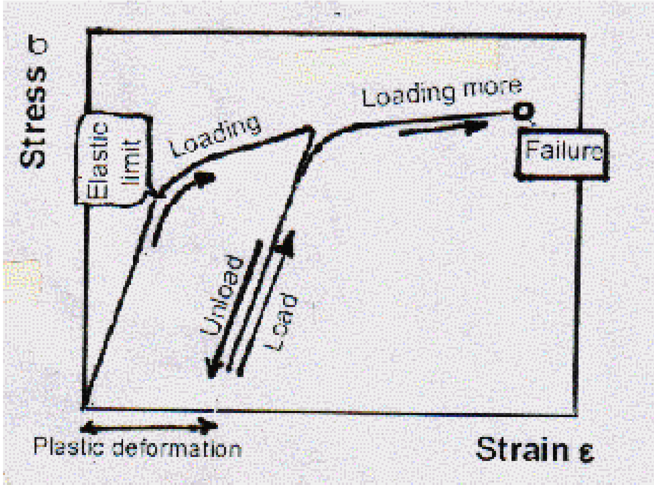
Strain **(£)** is a specific deformation, which is produced by the load applied to a body , or locally to the stress **(Q)** . When a load **(F)** is applied to a body with a length **(L)** this load produces a deformation **(dL).**

*** *The strain*** **(£)** is defined as the ratio between the deformation **(dL)** and the length **(L) :**

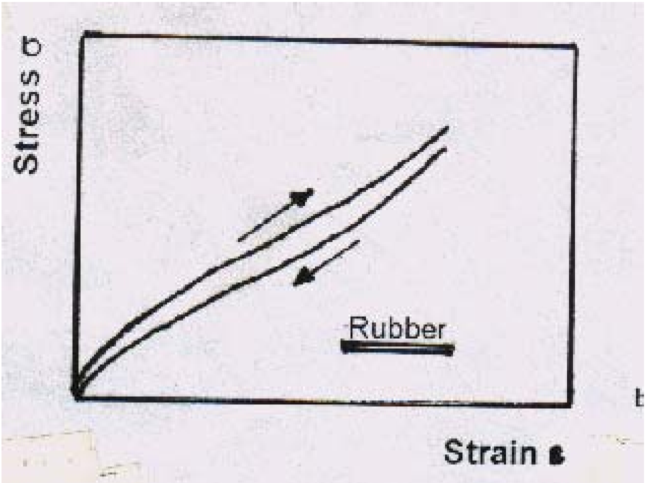
$$\text{£} = dL / L$$

The relationship between the applied stress and the resulting strain depends upon the materials concerned. Some materials, such as stainless steel, can be permanently deformed prior to failure, other are brittle, such as glass and also bone .

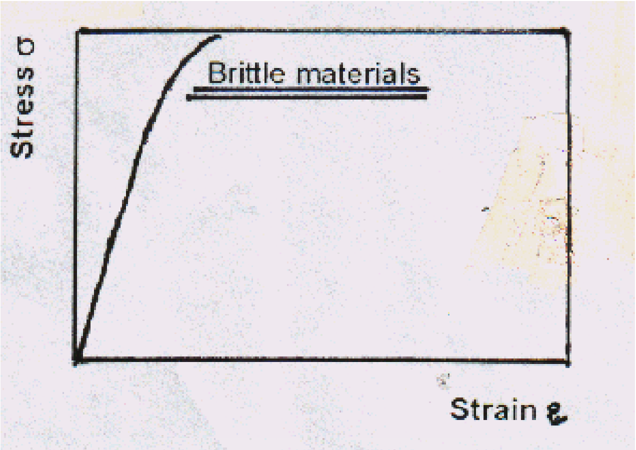
- There are materials, such as rubber or ligaments, for which the relationship between stress and strain rather complex and non- linear.



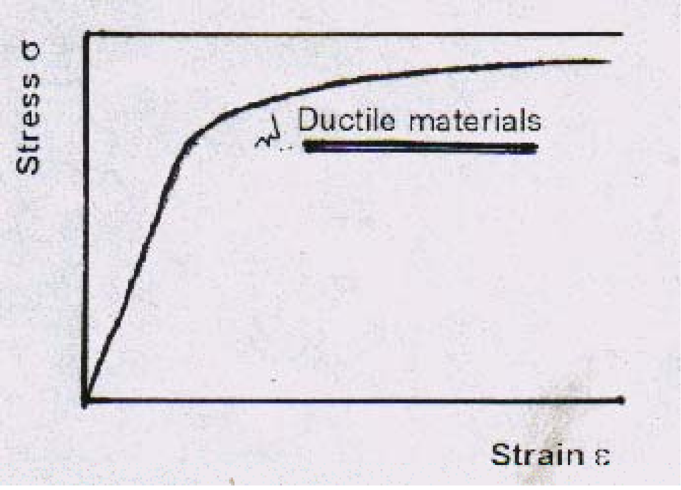
(Figure. a)



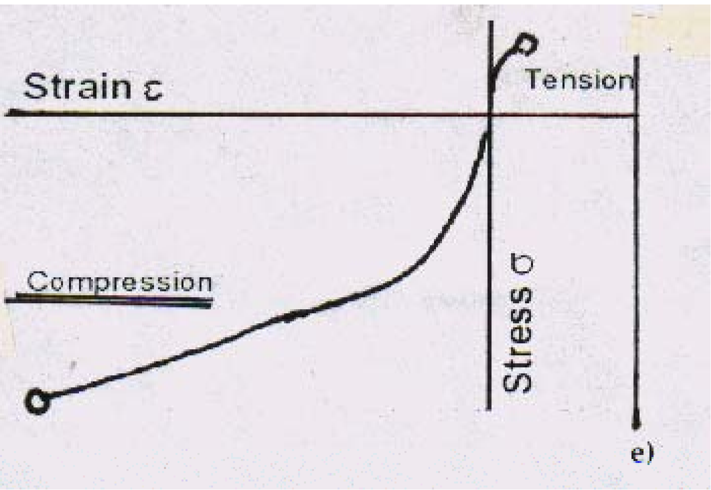
(Figure. b)



(Figure. c)



(Figure. d)



(Figure. e)

Relationship Between Stress and Strain for different materials : -

1) General relationship, **e.g.** for metals. Elastic and plastic range

2) Non- linear elastic **(i.e. Reversible)** relationship for rubber .

3) Brittle materials : very small plastic range, **e.g**. bone .

4) Ductile materials : Large plastic range, **e.g.** stainless steel .

5) For brittle material, the failure occurs at large values for compression than for tension.

- The most common relationship between stress and strain is shown in this **figure(a).**

For low values of stresses, the relationship between stress and strain is reversible and linear . This is the elastic range of deformation. When the stress is increased above a given value **(** the yield point, i.e. the elastic limit of stress **Q elastic)** , the strain increases more for the same increase of stress and the deformation is permanent .

This is the plastic range of deformation. When the stress is increased, it reaches a level of failure, the ultimate stress at failure **Q maximum** .

BONE : STRUCTURE AND MECHANICAL PROPERTIES :

**** *BONE*** is a rather complex material :

1) It is a biphasic material which consists of organic fibrous layers of collagen, including mineral microcrystals of hydroxyapatite .

2) Bone is dense along the diaphysis : cortical bone. At the metaphysis, cancellous bone shows a trabecular structure, surrounded by a thin shell of cortical bone .

3) Cortical bone has a microstructure, which might be lamellar, or Haversian**,** with a peculiar orientation of the collagen fibrils in the micro- lamellar structure of the lamellae or of the osteons. This orientation of the collagen fibrils and the density of the micro- crystals of mineral content **(Apatite)** affects the homogeneity of the cortical bone .

4) Cancellous Bone has a mechanical structure which was established last century by **Von Meyer and Cullmann and Popularized to Wolff.** More recent analysis has been by **Pauwels.**

- Bone is brittle for both tension and compression . It is somewhat stronger in compression, but bony materials show fairly symmetrical properties in tension and in compression.

  Cortical bone shows a fairly constant mineral density. It might vary slightly between different bones and different species. Furthermore, cortical bone is a reservoir of calcium and mineral density is not absolutely constant.

The Strain Gauges Used by ***Roberts*** in the mid **60 years** and propagated by ***Lanyon*** and ***Cochran*** in early **70 years** are still **(**with minor modification**)** the strain gauges we will be using in the ***Year 2000.***

THE GAUGE IS ATTACHED TO THE BONE AS FOLLOWS :

1) Minimal stripping of the periosteum in the area to which the gauge will beApplied .

2) Remove the fat with chloroform and wet the bone using sterile water .

3) A small amount of adhesive **(Cyanoacrylate)** is applied to the undersurface of the gauge.

4) …. Which is then immediately placed onto the prepared surface of the bone being careful to observe correct alignment .

5) Teflon foil is placed over the gauge which is pressed onto the bone for about **30 seconds**. The foil prevents the surgical glove from sticking to the bone .

6) The teflon foil is carefully removed without damaging the gauge. The gauge is now ready for use. To be absolutely sure that the gauge will not slip a drop of cyanoacrylate is often added to the top of the gauge and left to set for about one minute.

Complications :

1) Firstly, Gauges may become loose .

! How long does a strain gauge remain stuck to the bone ?

* If the appropriate adhesive is used, i.e. an adhesive which will not dissolve in water or biological medium, the limit depends on remodelling on the undersurface of the bone. The start of remodelling and some resorption has been observed at **12 weeks** on the sheep tibia. However, I would not recommend using this strain gauge technique for longer than **5 – 6 weeks** .

2) The Second complication is the disparity of insulation resistance over time. Leakage may occur, sometimes because the wires outside the skin have become moist and lead to incorrect data .

3) The Third complication is the breakage of the connecting wires. This has been the reason why we installed a **5 cm diameter** loop next to the gauge .

Some Notes About The Strain Gauges :

- Small electrical resistors.
- Positive deformation is traction.
- Negative deformation is compresion.
- The gauge should always be checked before implantation.
- The band extensometry to be submitted to an effort sufers a deformation.
- The gauges only need be left for **24 hours** to dry thoroughly for their insulation resistance to reach an adequate level.

The Strain Gauge Technique is inexpensive**:** The basic instrumentation **(** preparation instruments, basic measuring equipment, computer for data analysis **)** might be estimated at about **20.000 €** i.e. **€ 2000** for the gauge preparation tools, **€ 8000** for the bridge amplifiers and strain gauge testers, **€ 8000** for the personal computer including data analysis software, **€ 2000** for a basic set of strain gauge, coating etc.

METHODS.

METHODS.

The Knees were dissected carefully leaving the Patellofemoral Joint, Tibiofemoral Joint, the periarticular skin and subcutaneous tissue, and the knee capsule intact. The rectus femoris and quadriceps tendons were left for clamping. Extra care was taken to ensure that no cartilage or ligament damage occurred during dissection. The tibia and femur were each sectioned transversely at a point **Twenty-Five Centimeters** from the joint space and were mounted in knee testing system.

The specimens was then clamped rigidly into steel cylinders in its anatomic position by aligning the cylinder along the axis of the proximal tibia and distal femur. Rigid fixation of the specimens was achieved by using 12 positioning pins in additionto two bolts driven through the diaphysis of the Femur, Tibia, and Fibula.

The quadriceps tendon was wrapped with gauze and clamped at a point five centimeters proximal to the proximal pole of the patella, allowing knee-flexion angles of as much as 120 degrees without interference with the femoral groove. With the additional length provided by the fixtures, the resulting functional lengths of both the tibia and femur were approximately thirty-five centimeters.

The femoral-tibial clamping system was designed to hold the specimens at a specified angle of knee flexion. The specimens were mounted in a testing ring. The testing ring allowed simulation of different values of femoral neck anteversion and tibial torsion.

Six-Extensometric Gauge (Strain Gauge) were fixed in each one of the patella, knee flexion was performed applying a force on its mechanical axis. Measurement obtained by the gauge were recorded dynamically as microdeformities every degree of the knee.

Preparing and Keeping the Specimen:-

First of all, the limb is unfrozen at room temperature, with an unfreezing delay of 10 hours approximately. Once the specimen is unfrozen the distal end of the tibia is fixed in a circular Polyurethane cubeta, leaving a distance of 24 cm from the articular interlineto the distal end of the tibia and 10 cm for Femur superior. Fixation or inclusion is reached through an industrial synthetic Polymer **(Polymerid).** It is absolutely necessary to keep a perfect and absolute perpendicularity between Tibia and holding cubeta. **Show in this figure**

Image in which we can appreciate the inclusion of the specimen on the distal polyurethane cubeta. This fixation allows performing some tests without tiredness elements on the distal end of the specimen.

Once the specimens was securely mounted and positioned at a desired knee flexion, the neutral position of the patellofemoral joint was defined and determined as the position that exhibited minimum tension in the quadriceps tendon. The correct determination of the neutral position was essential for accurate assessment of the patellofemoral contact pressure distribution for each respective specimen.

We had done **6 holes** on the anterior surface of the patella as following:-

3 holes lateral and 3 holes medial, we use the strain gauges to measure the accurate values for the stress. The procedure for this process will be at the following, we drill a hole into the top view surface of the patella. The length of the hole relies on the thickness of patella. We start from the top surface and go deeply; however, before we reach the bottom surface by **3 mm,** we have to stop to implement the best result for the stress.

e.g. If The Thickness Of Patella = 20 mm, The Length Of The Hole = 17 mm And So On …!!

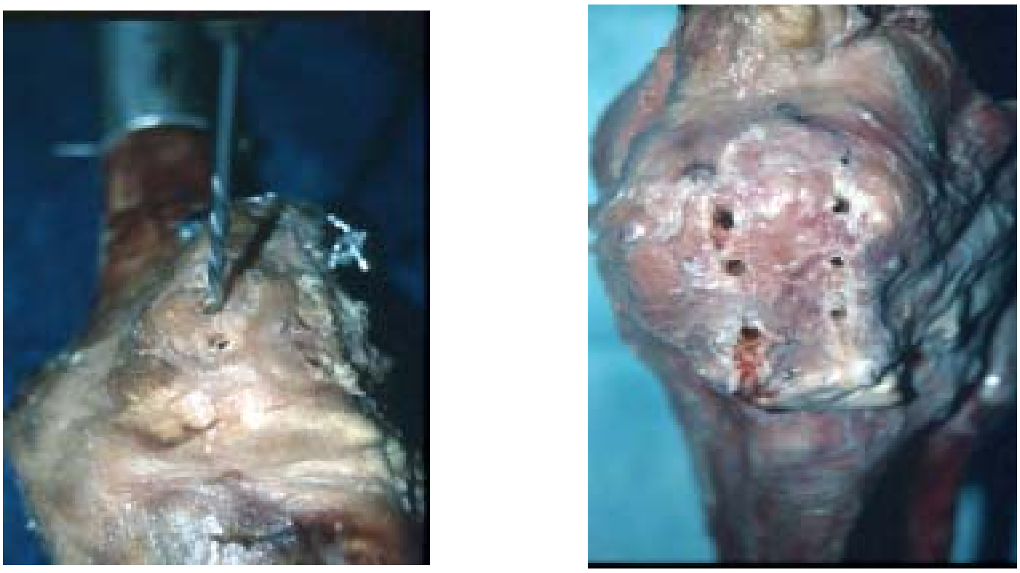


Figure 1: Image For 6 Holes To Pass The Strain Gauges Through Its.

And after then: -

1) Minimal stripping of the periosteum in the area to which the gauge will be Applied .

2) Remove the fat with **Chloroform** and wet the bone using sterile water .

3) A small amount of adhesive **(Cyanoacrylate)** is applied for both the undersurface of the gauge and the hole, immediately a Strain Gauge has been put in each hole.

4) …. Which is then immediately placed onto the prepared surface of the bone being careful to observe correct alignment .

5) Teflon foil is placed over the Gauge which is pressed onto the bone for about **30 Seconds**. The foil prevents the surgical glove from sticking to the bone .

6) The teflon foil is carefully removed without damaging the Gauge. The gauge is now ready for use. To be absolutely sure that the Gauge will not slip a drop of Cyanoacrylate is often added to the top of the Gauge and left to set for about one minute.

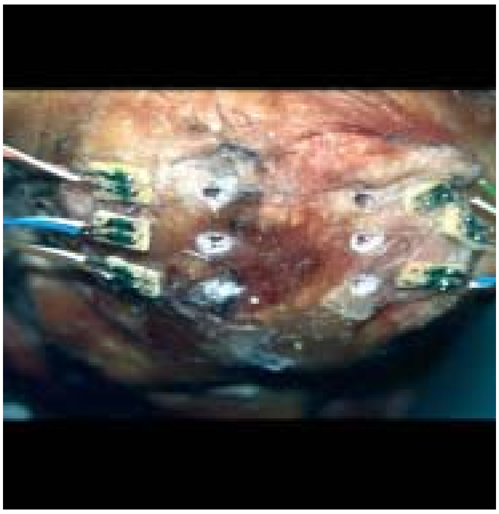


Figure 2: Image For 6 Strain Gauges Attached With The Anterior Surface Of The Patella.

Figure 3: **Anterior View Of The knee With Strain Gauges.**

Figure 4: **Anterior View Of The Knee With One Ring Attatched With Quadriceps Tendone And Another Ring Attached With Femoral Biceps Tendon.**

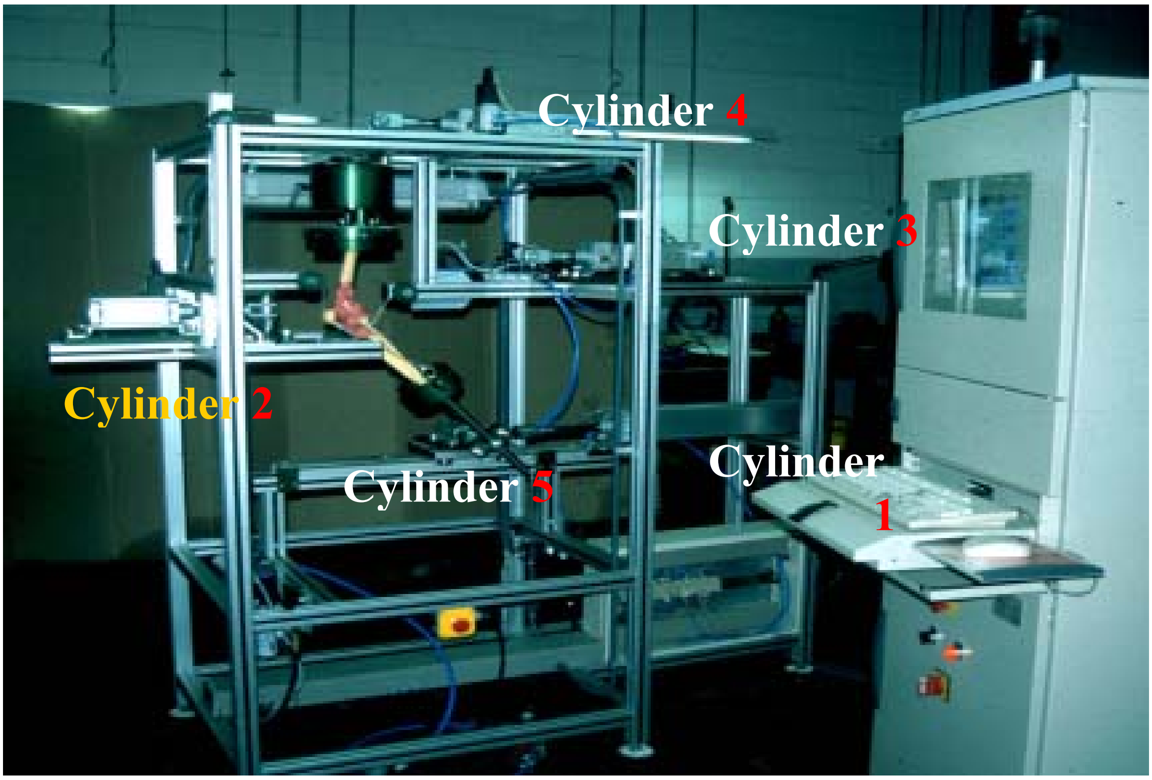


Figure 5 : **Simulator (Type) II . With 5 Hydropneumatic Cylinders.**

Figure 6: **Quadriceps Tendon.**

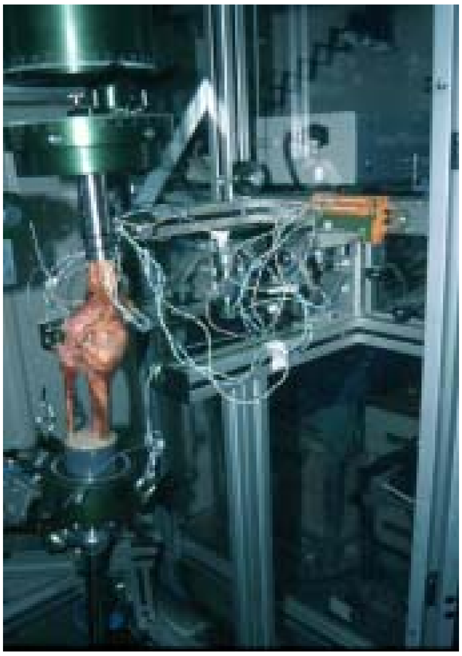


Figure 7: **The Specimen Fixed To The Special Device And Was Clamped Rigidly Into Steel Cylinders.**

Figure 8 : Image Of The System Of Record Channels.

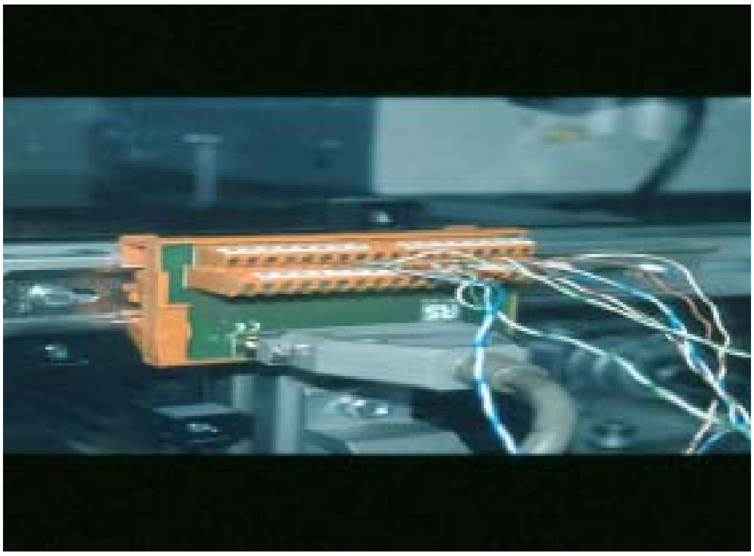


Figure 9 : Image Of The Connect Of The Cables.

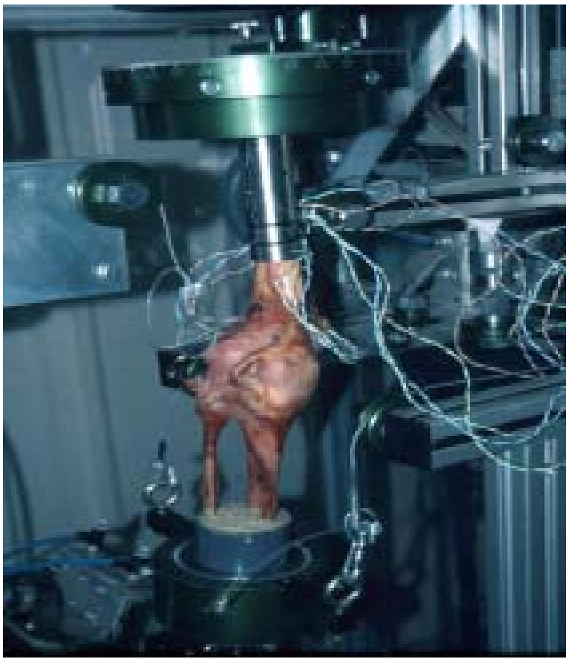


Figure 10 : Picture Of The Double-Cylinder System That Enables Modification Of Femoral Anteversion.

Above picture shows that we have used a head graduated cylinder, which enable us to place the legs in different positions by controlling the angles. In the meantime, we have changed the position by adjust the angle from **10^0 to 45^0** . In each stage, we have taken five readings for the same angle to achieve the best results, which could accomplish our target of this ***Scientific Research***.

Figure 11 : (Picture it is observed the assembly of the proximal femur on the system of two proximal hoops that simulate wing femoral anteversion. The system proximal possesses great inflexibility so that is not alterations in the efforts upon accomplishing the flexo-extension) .

After it is accomplished the setting of the system on the tibial lace. For the place, put of the hoop are introduced two nails of **Steinmam of 4mm** cross to **90^0** through the orifices that the hoop possesses in all its periphery. Upon putting the hoop is indispensable that its.

Distal surface remain perpendicular with the shaft of the femur, and that the proximal expansion of the hoop is located in the internal side place that to occupy to the head of the femoral in a normal extremity.

Figure 12 : **(Above Picture Shows That We have used to control the degrees it is the assembly of the system of inclusion on the distal platform that also of anteversion)**

We have put the specimen to **10^0** femoral anteversion. It is accompished modifying the position of the proximal expansion for the system of cylinder concentric. Is is begun the work routine in the one which, each one of the cylinders simulates an effort of the cycle of the step. They are accomplished records extensiometry during **5 Consecutive Cycles**. **(Image 12)**.

Figure 13 : knee During The Flexion (The Initial Position). Attached With Extensometries.

And also this **(Image 12).** Is observed the specimen during the flexion and all the system of connexion with Extensometries.

NOTE :-

- **STRAIN** = Length – Elongation Length / Length.
- **STRESS** = Forces / Area.

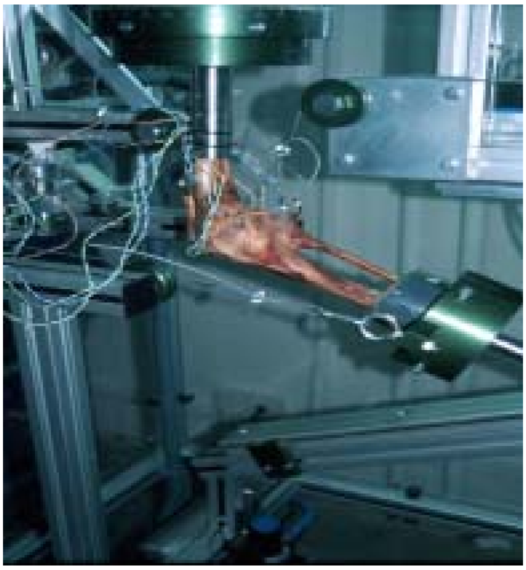


Figure 14 : The Picture Shows The Knee With A Immense Flexion.Posterio View Of The Specimen.

Figure 15: Picture Of The Simulator Control System Composed By A Control Station.

All the records Extensometries are collected through some point that permit the movement of the data to **Excel Program** in the one which be accomplished the caculate.

Figure 16 : Picture Of The Simulator Control System With Computer System.