WHAT IS BIOMECHANICS ?

Biomechanics is mechanics applied to biology

Mechanics usually describes

- force, motion, strength of materials, etc.

interrelation also with, as examples

- physics of atoms, molecules, etc. -> **quantum mechanics**
- physics of solids, liquids, gases, and phase transitions-> **thermodynamics and statistical mechanics**
- physics of living systems->**biomechanics**

Biomechanics investigates the mechanics of living systems, i.e. biological components and structures

That includes the investigation of

- the normal function
- alterations due to maturation and ageing
- alterations due to exercise
- alterations due to diseases
- helps to design artificial intervention

Thus biomechanics is also associated to

Diagnosis, Surgery, Prostesis, etc.

- design implants
- how to apply screws in surgery
- and also anchors to fix tendons to bone

As to the investigation of components of living systems; here is a list

- bones
- articular cartilage
- ligaments
- tendons
- (palmar) aponeurosis and especially the „uniaxial" rays to the fingers; with reference to DUPUYTREN´S CONTRACTURE
- samples of skin; variations of structure and biomechanics with body region and posture; with reference to the LANGERS CLEAVAGE LINES.

This lecture concentrates on the **biomechanics of soft connective tissues**

i) Biomechanics of **parallel fibered tissues**

- stress-strain relationship
- tensile strength characteristics
- viscoelasticity and plasticity

ii) Biomechanics of the **most prominent components** of the parallel fibered tissues

- collagen network
- elastin network
- ground substance matrix

iii) Viscoelastic modelling of the mechanical properties of biological tissues

iv) Biomechanics of **flat tissues**

- palmar aponeurosis
- human skin

v) Remarks on **experimental techniques**

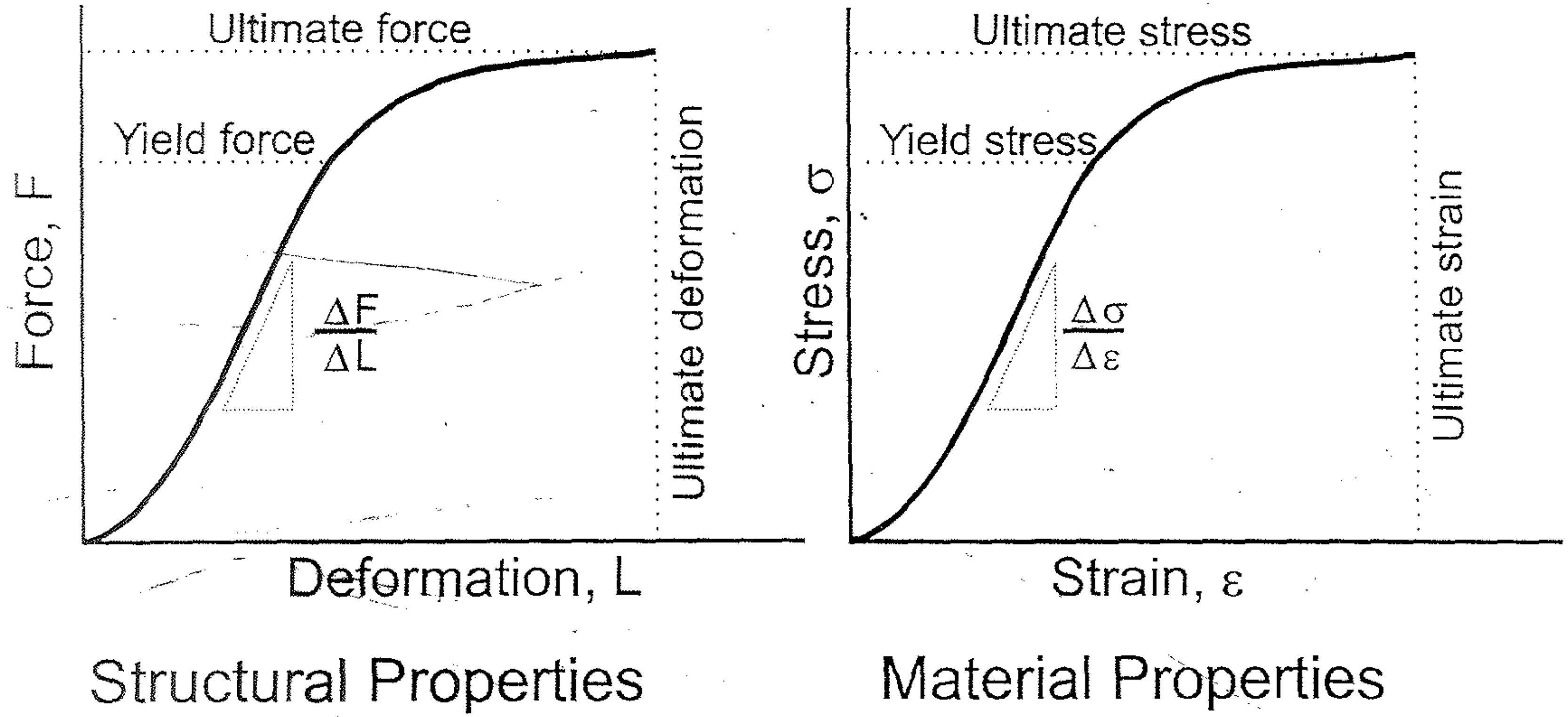


Figure 3 - Force-deformation and Stress-strain curves illustrating structural and material properties respectively.

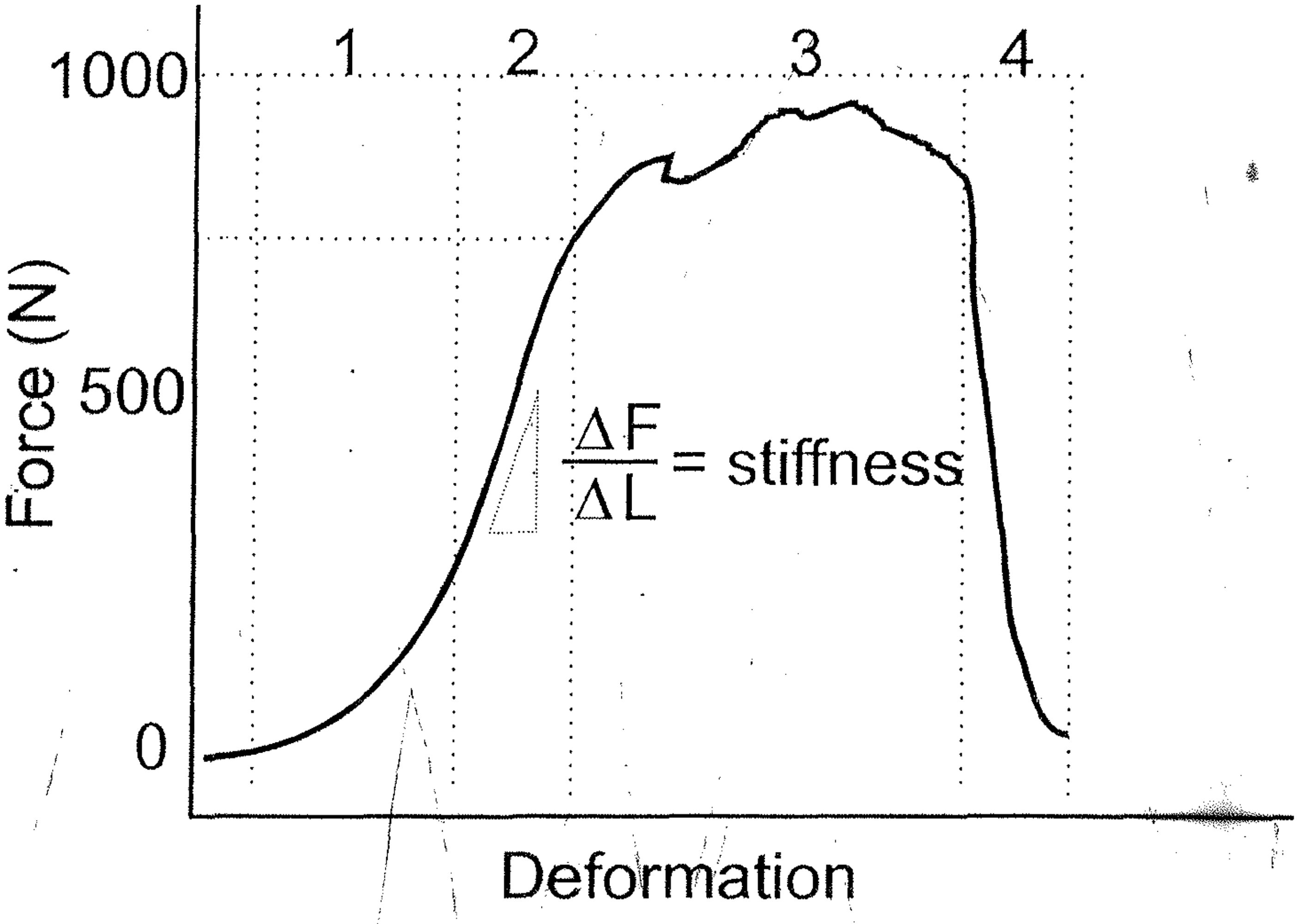


Figure 3 -A force-elongation curve obtained from a tensile test to failure of a rhesus mo femur-ACL-tibia preparation is shown. There are four regions that are commonly used describe a force-elongation or stress-strain curve. Region 1 is termed the "toe region" a a non-linear increase in load as the tissue elongates. Region 2 represents the linear regi curve. In Region 3, isolated collagen fibers are disrupted and begin to fail. In Region 4 ligament completely ruptures. (Modified from Butler, D.L., Grood, E.S., Noyes, F.R., a Zernicke, R.F., Biomechanics of ligaments and tendons. Exercise and Sports Science R 6, 125-181, 1984.)

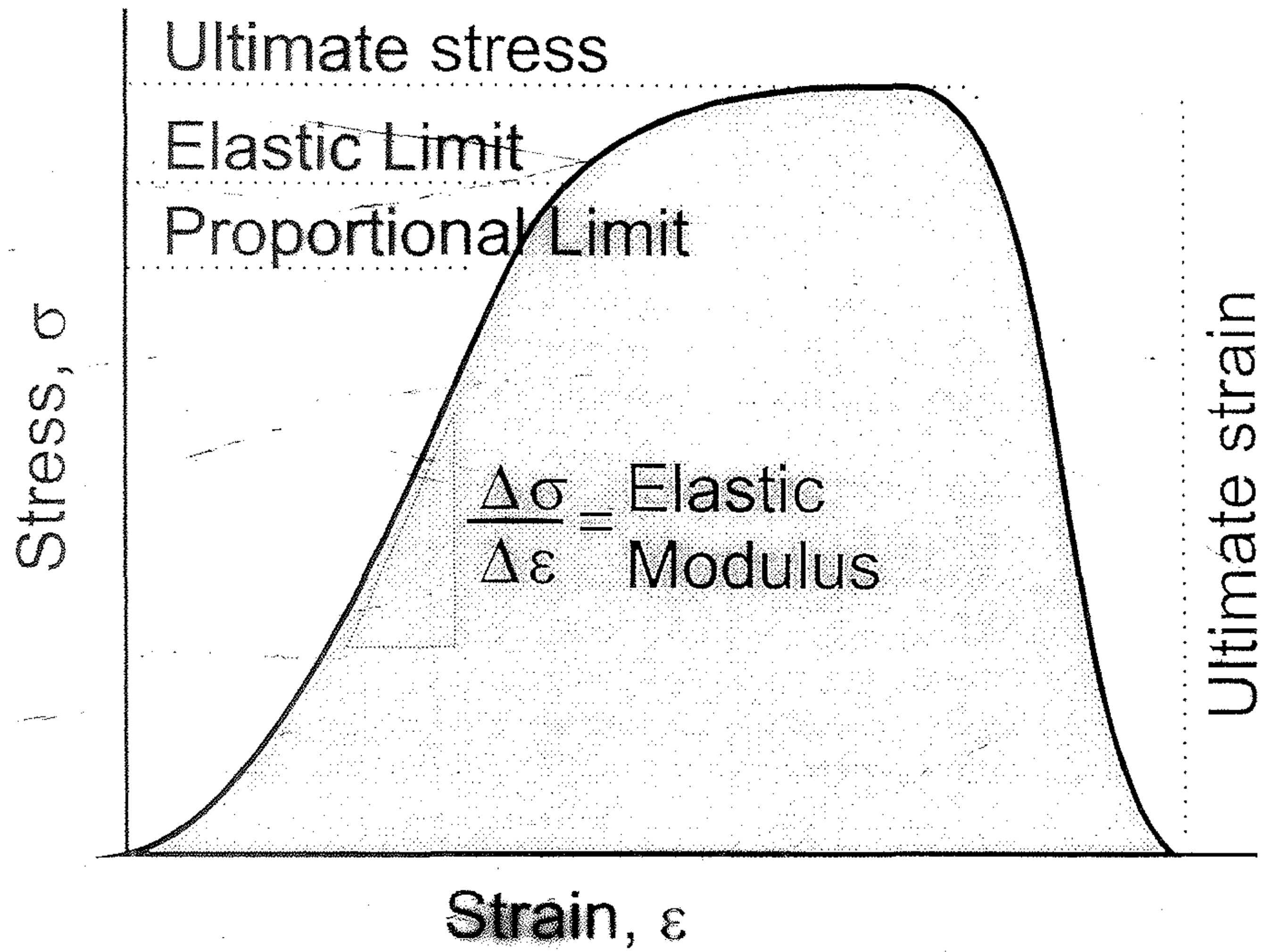


Figure 4 - Illustration of a stress-strain curve with various quantities of interest identified.

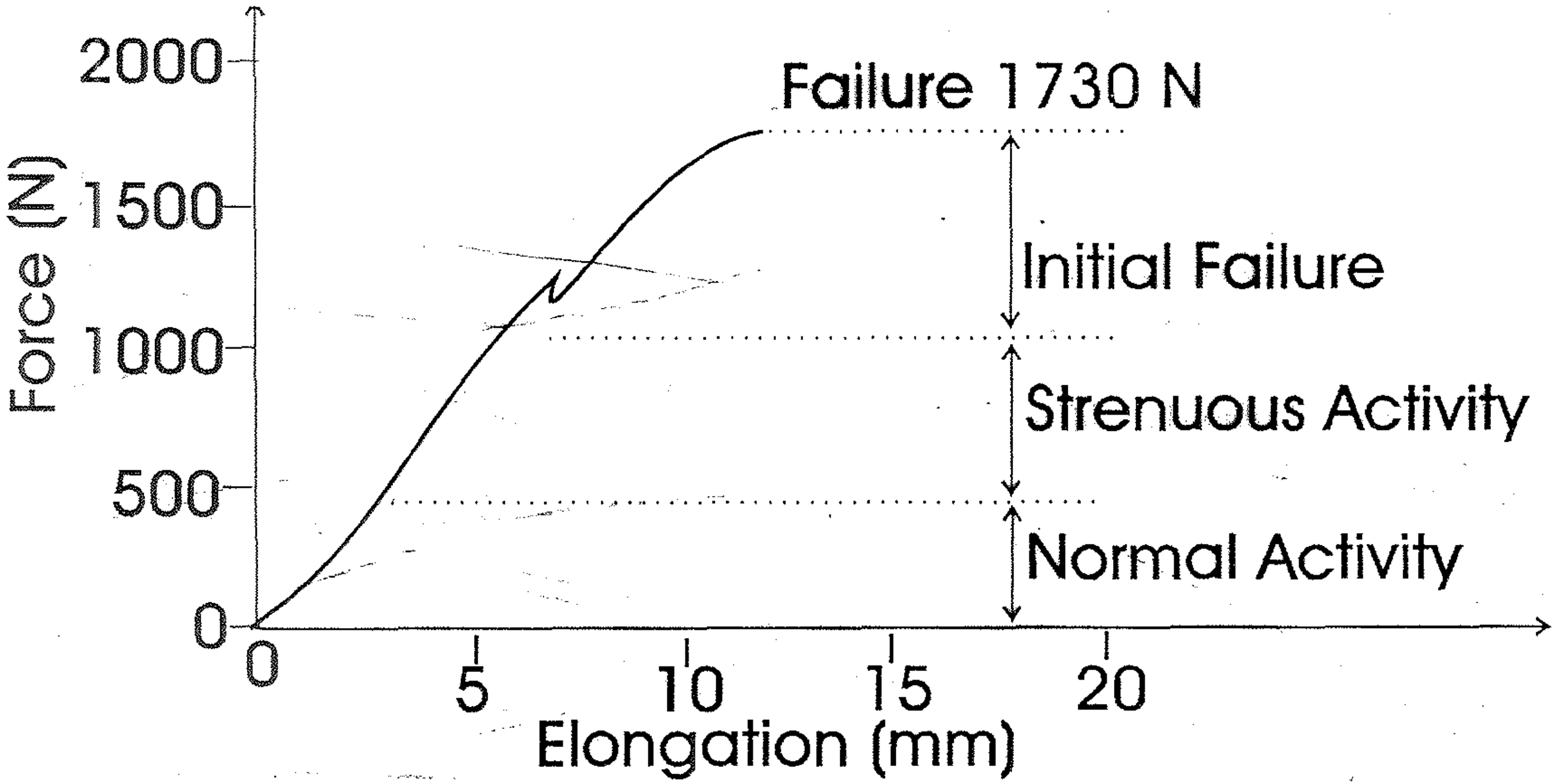


Figure 5 -A hypothetical force-elongation curve for a human ACL-bóne complex is illustrated along with daily activities that correspond to specific loading levels. During routine daily activities such as walking and standing, ligaments are loaded to less than one fourth their ultimate tensile load. During strenuous activities such as fast cutting during intense running, loading levels may enter into region 3 where isolated fiber damage takes place. (Modified from Noyes, F.R., Butler, D.L., Grood, E.S., Zernicke, R.F., and Hefzy, M.S., Biomechanical analysis of human ligament grafts used in knee ligament repair and reconstruction. Journal of Bone and Joint Surgery, 66A(3), 344-352, 1984.)

Some conclusions from the diagrams shown

The diagrams showed typical biomechanical properties of tendons and ligaments, such as

i) Tendons show a high compliance at low strains, i.e. in the range of **1% to 3% of strain**. This is the „toe" region, characterized by incresing stiffness

ii) Tendons show a lower compliance in the range of strain of **3% to 5%.** Linear range of the stress-strain graph; usually stiffness is computed there; it is the range with steepest ascent

iii) Failure will occur at **8% up to 15% of strain**

The stress- strain (or force-elongation) relationship is nonlinear. Obviously the ligaments or tendons allow movability (limbs) in normal use („toe" region, beginning of linear range), and lock these movabilities to prevent from overuse (end of linear range).

Contribution to Stress-strain properties and arrangement of the main constituents

i) **collagen** (mainly type I) is the **main load bearing component** and contributes about 70% of dry weigth (dependent on the type of tendon or ligament)

ii) **elastin** contributes mainly the the stress-strain properties **in the toe-part**, i.e. in the range of low strains; 1-2% of dry weigth of the tissue, except some tissues like lig. Nuchae or lig. Flavum

iii) **proteoglycan matrix** (ground substance matrix) which can be seen as a medium allowing the tissue fibers to **glide past each other** when strain is applied („lubricant")

Relation of the stress-strain properties to the hierachical structure of ligaments

Compound of the **networks of collagen and elastin**

These networks are embedded in the **ground substance matrix**

The hierarchy may be described from the molecular level to the macrocopic level as

- collagen triple helix
- five of then form the fibril
- subfibril
- fibril
- fibre
- fascicle with the typical crimp pattern
- ligament

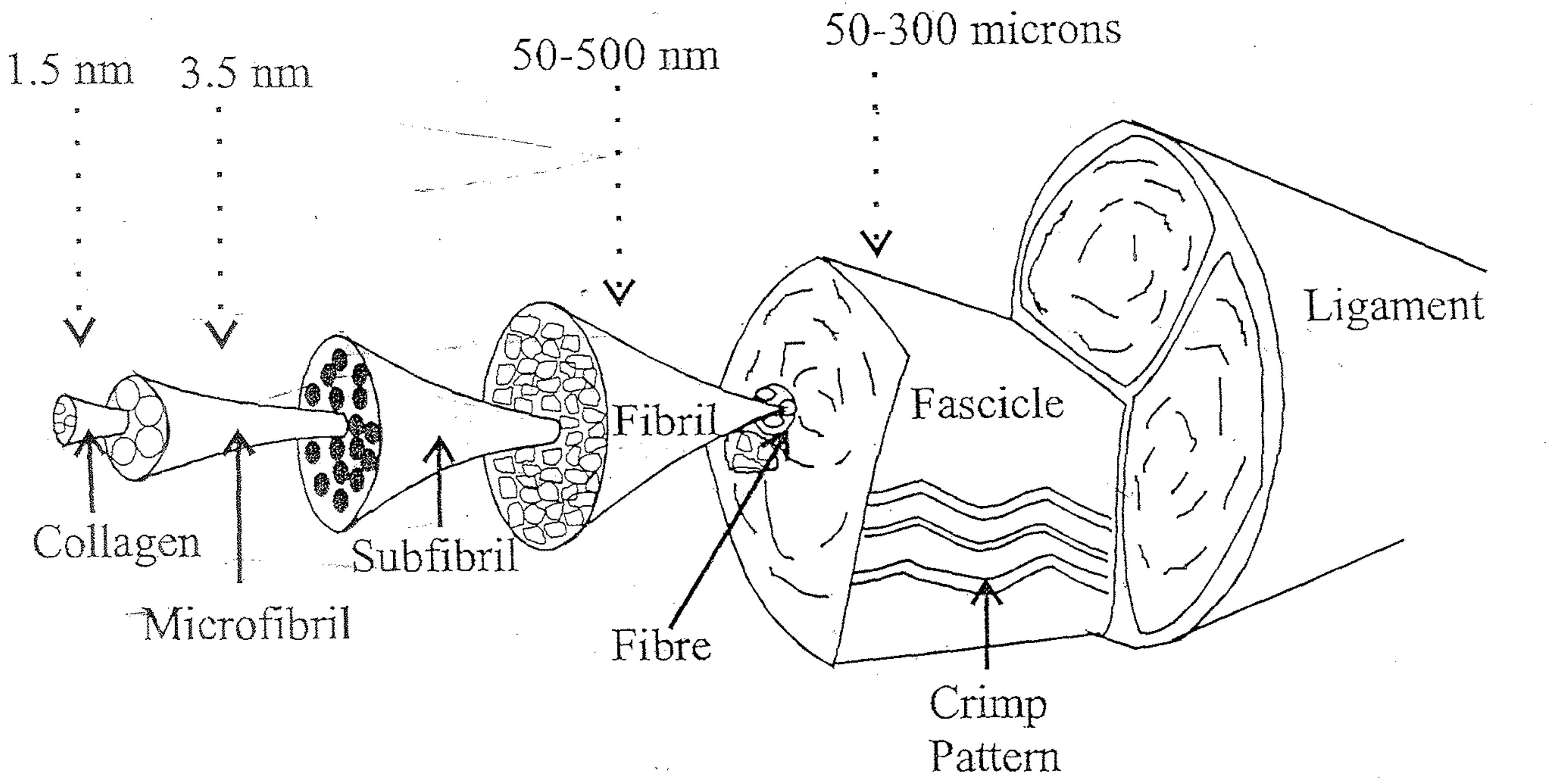


Figure 1 -A schematic diagram of the structural hierarchy of ligament is shown. L
composed of smaller and smaller fiber bundles. The basic structural element is the
molecule.

Changes of the hierarchical structure due to application of strain

- toe part. At low strains the **crimp pattern** of then subfibril **disappears** with straining as can be monitored with polarization microscopy
- **alignement** of the **staggered arranged collagen molecules** (analysis by X-rays) at the beginning of the linear portion of the stress-strain relationship
- **collagen molecules glide past each other** in the linear portion (also observed by X-rays)

Even at 10% of strain of the tendon or ligament the molecule itself may be elongated olny be 1% strain.

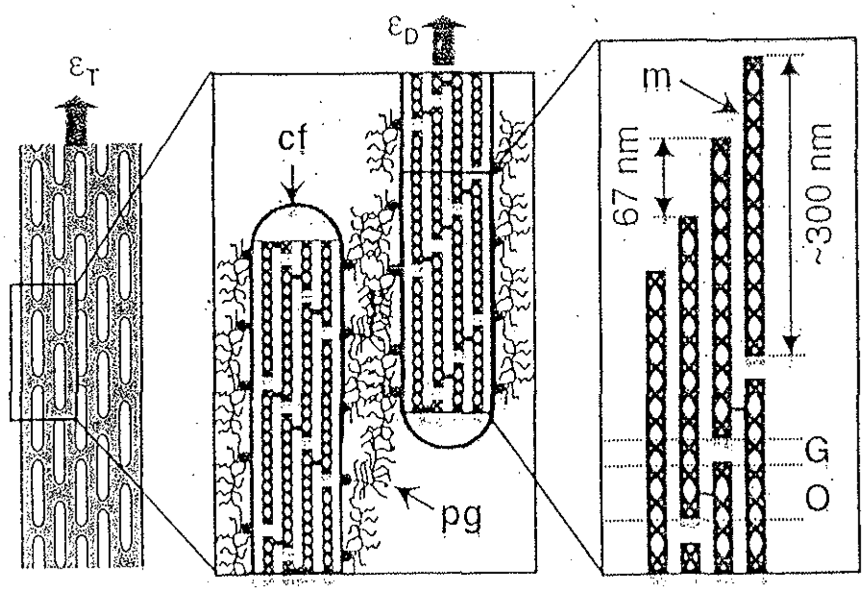


Figure 9.2: Simplified tendon structure (see, e.g., Vincent, 1990). A tendon is made of a number of parallel fascicles (a) containing collagen fibrils (b). The fibrils (cf) are typically coated with proteoglycans (pg). They have a thickness of several hundred nanometres and a length in the order of 10 micrometres. Triple-helical collagen molecules (m) are packed within fibrils (c) in a staggered way with an axial spacing of $D = 67$ nm, when there is no load on the tendon (Hodge and Petruska, 1963). Since the length of the molecules (300 nm) is not an integer multiple of the staggering period, there is a succession of gap (G) and overlap (O) zones. The lateral spacing of the molecules is around 1.5 nm, but the full three-dimensional arrangement is not yet fully clarified (Hulmes et al., 1995; Wess et al., 1998). ε_T is the total strain of the tendon and ε_D the strain in the fibrils, which can be measured via the change in the axial D-period due to tensile loading.

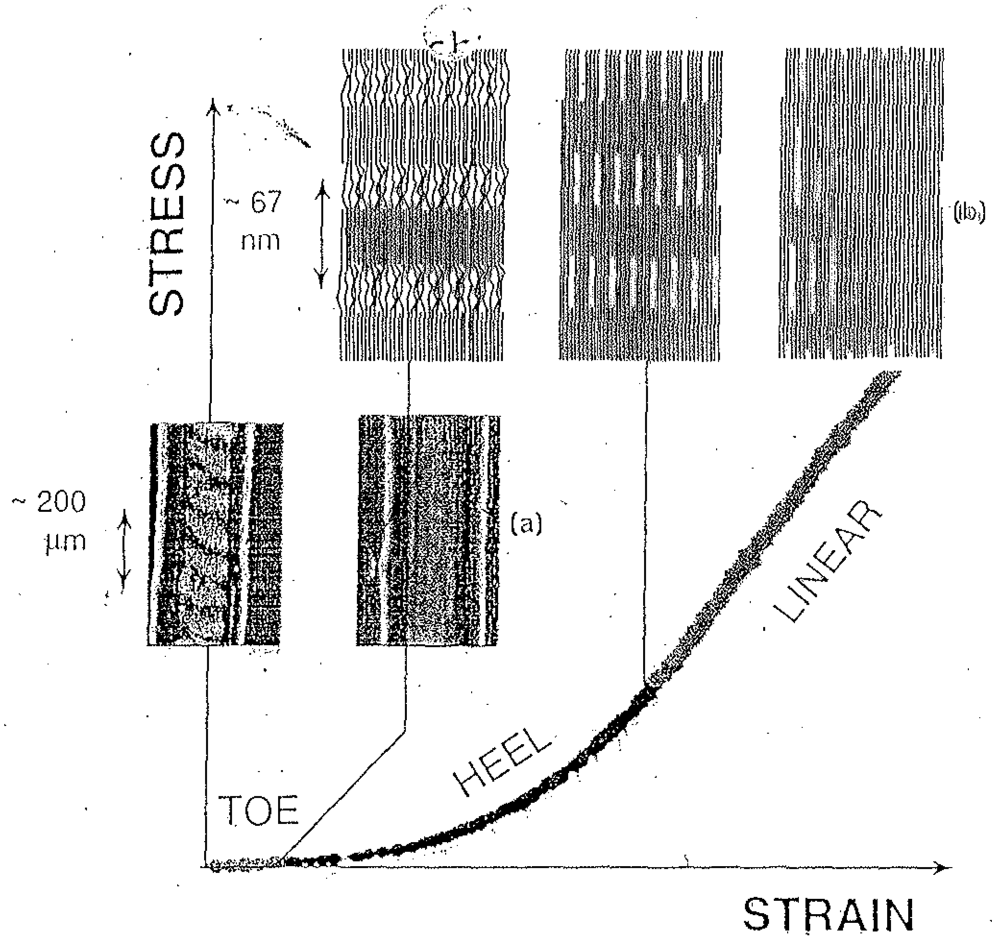


Figure 9.3: Schematic behavior of the normal collagen fibril structure from rat-tail tendon (marked cf in Figure 9.5) during tensile deformation. From 'Fibrillar Structure and Mechanical Properties of Collagen', by P. Fratzl et al., 1997, *J. Struct. Biol.* 122, 119–22. The experiment was performed at a strain rate where the actual strain of the fibril (ε_D) was about 40% of the total stain of the tendon (ε_T) in the linear region. Plotted on the horizontal axis is the total strain of the tendon (ε_T). Previous synchrotron X-ray scattering experiments (Misof et al., 1997) have shown that the tendon structure goes through a sequence of changes upon stretching. First, a macroscopic crimp in the tendon (Diamant et al., 1972) is straightened out, as visible in the polarised light (a). Then, microscopic kinks in the collagen molecules (located mostly in the gap region of the fibril structure) are removed, leading to an entropic contribution to elasticity (b, left). Finally, the molecules start to glide past each other in the linear region of the stress-strain curve (b, right).

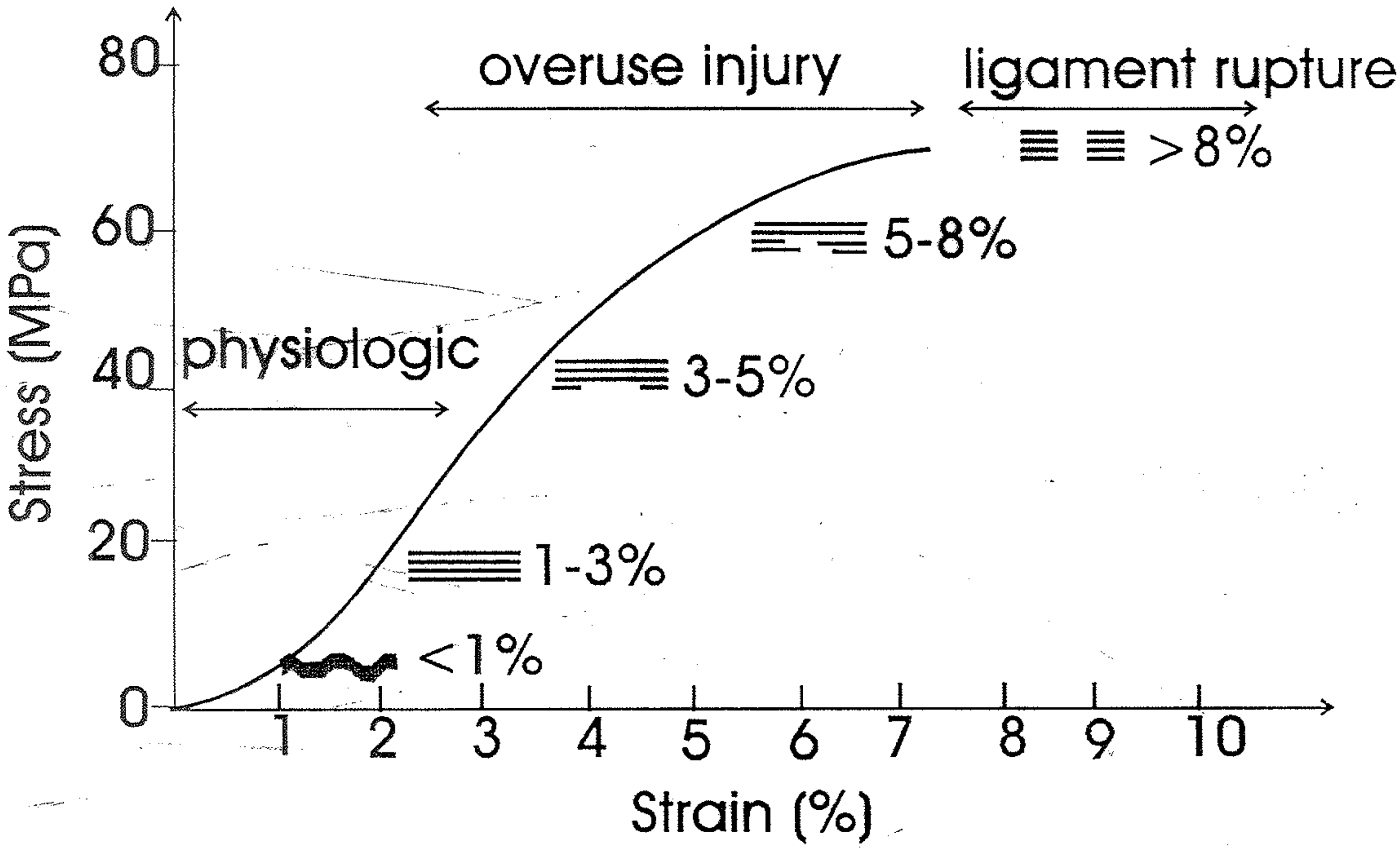


Figure 4 -A stress-strain curve illustrating the relationship between changes in the collagen "crimp" pattern, or stretch, and ligament mechanical properties is shown. Increases in ligament strain in the "toe region" of the curve results in straightening of the "crimp" pattern. During the linear portion of the curve the collagen fibers are stretched. As the ligament is further strained isolated ligament fibers begin to rupture and if deformation continues, then complete ligament fail occurs. (Modified from Butler, D.L., Grood, E.S., Noyes, F.R., and Zernicke, R.F., Biomechanics of ligaments and tendons. <u>Exercise and Sports Science Reviews</u>, <u>6</u>, 125-181, 1984.)

Conclusion. Mechanisms of deformation of tendons

(i) deformation of the collagen molcule

the molecule is stable and shows small deformation only

the deformation may be described as an increase of the gap between the staggered arranged molecules

gliding of molecules past one another

(ii) the deformation of the fibril >> then (i)

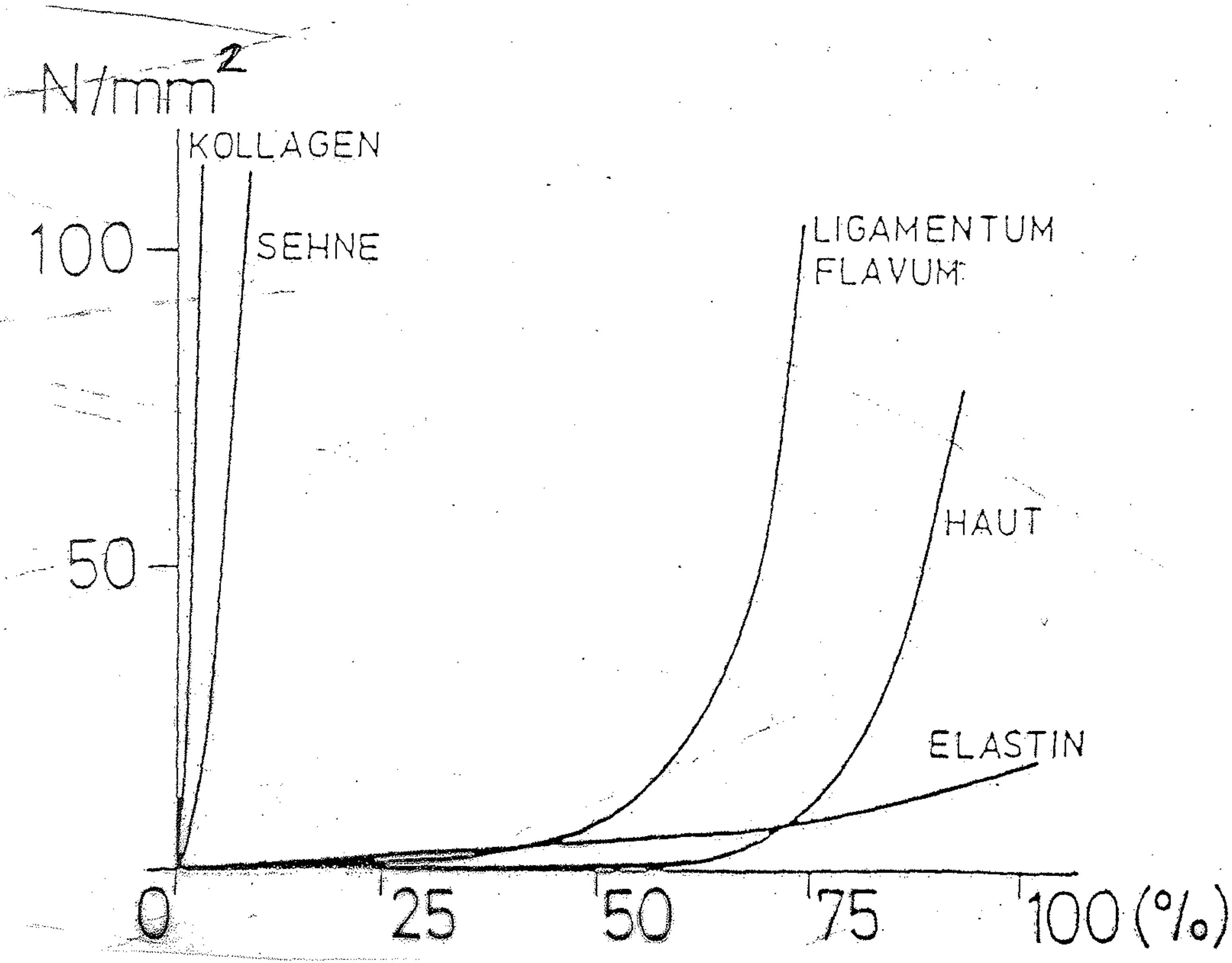
(iii) the deformation of the tendon >> then (ii)

Comparison of the stress-strain relationships of the tissue components collagen and elastin with those of tendon and skin.

Tissue components may be singled out from the tissue by enzymatic degradation of the other components

As seen on the following graph stiffness decreases in the order

- Collagen
- Tendon
- Ligamentum flavum (elastin content larger than in other ligaments)
- Skin (dependent on direction of loading due to anisotropy)
- Elastin


N/mm²
KOLLAGEN
SEHNE
LIGAMENTUM FLAVUM
HAUT
ELASTIN
100
50
0
25
50
75
100 (%)

We concluded that the nonlinearity of the stress-strain diagram may be attributed to

- physiological function of tendons and ligaments
- the hierarchical arrangement of diverse substructures of the tissues

There are further properties to be taken into account for experiments with bio-tissues. **The stress is composed of time-independent (elastic) and time-dependent (viscous and plastic), i.e. strain rate dependent contributions.**

- the stress-strain relationship depends highly on the **strain rate**
- the loading and unloading branches are different resulting in a **hysteresis loop**
- if the load is kept constant the strain will increase **(creep phenomenon)**
- if the strain is kept constant the stress will decease **(stress relaxation)**

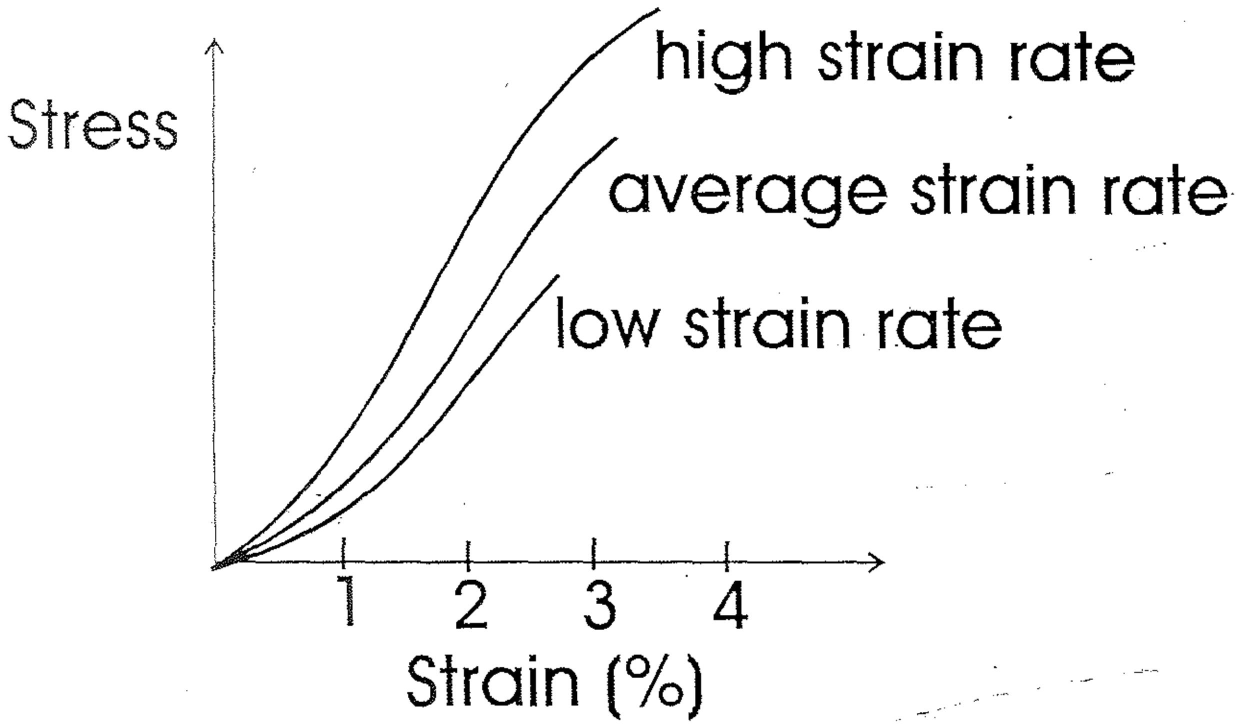


Figure 8 - Strain-rate affects on stress-strain properties. The faster a tissue is loaded the greater the stiffness and the higher the ultimate failure load. Some tissues may exhibit higher ultimate strains for higher strain rates (e.g. ligaments), others may have lower ultimate strains (e.g. bone).

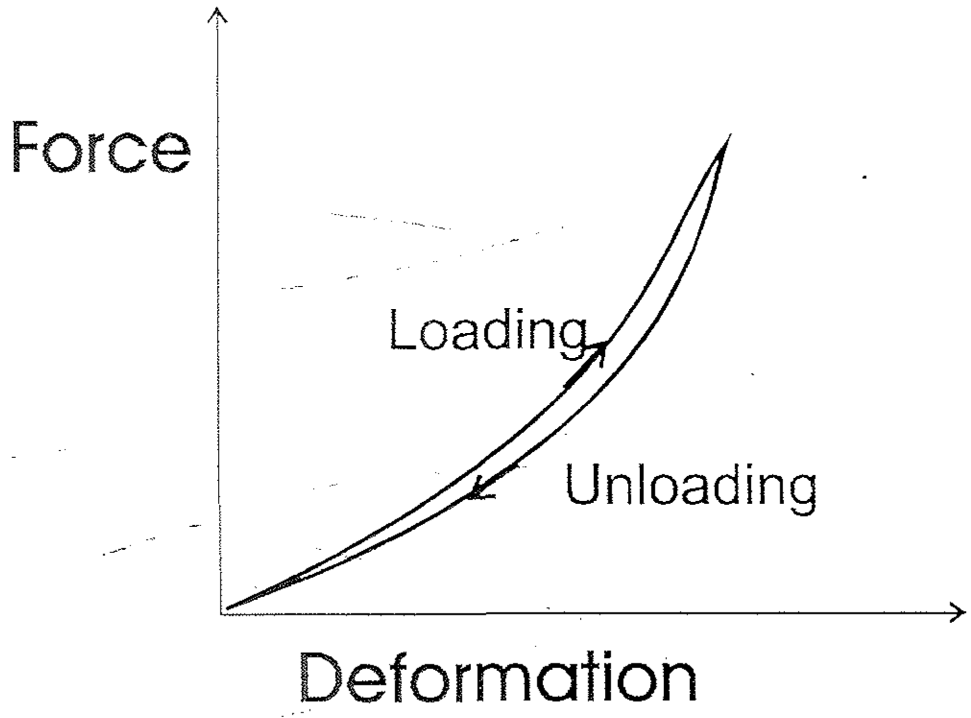


Figure 9 - Loading and unloading curves during a tensile test of a biological tissue. The two curves are not identical. The difference in area under the two curves is the area of hysteresis and represents the energy lost due to internal friction within the material.

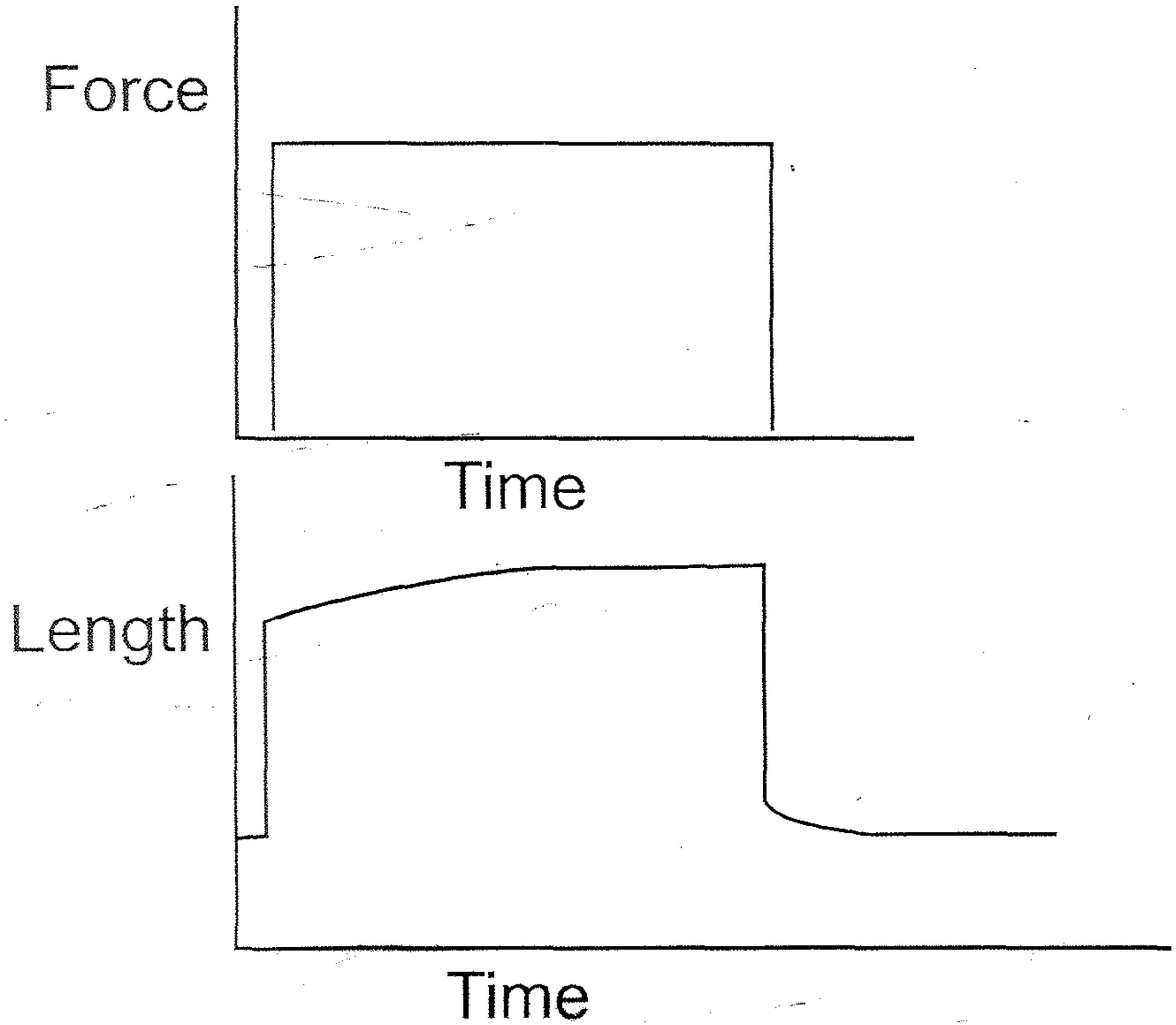


Figure 7 - Example of a creep response. Such a response might occur in a tendon subjected to a constant muscle force.

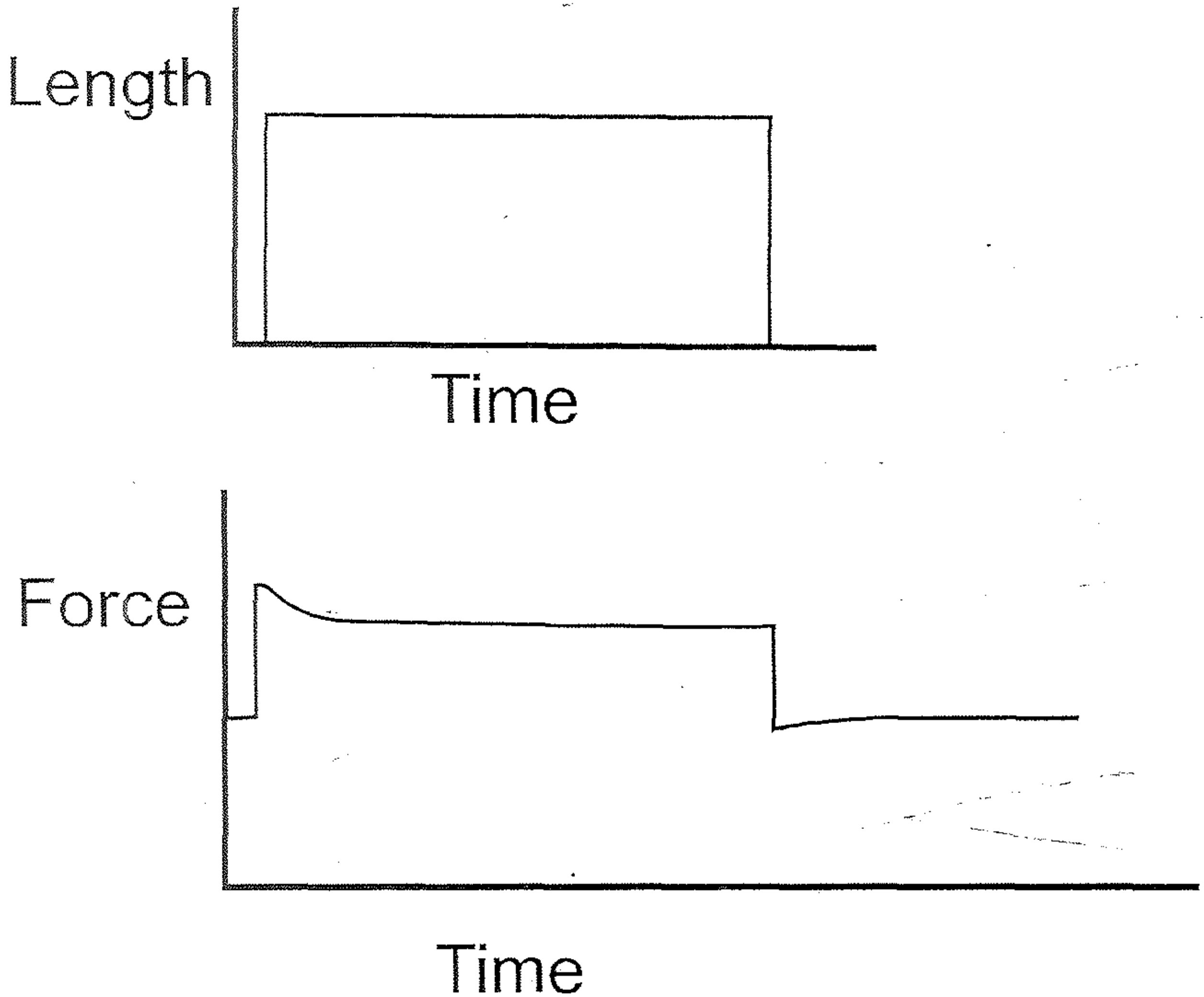


Figure 6 - Example of a force or stress relaxation response. Such a response might occur in tissues around the knee if the knee is moved to a specific knee position and maintained in that position.

As to a conclusion

- the higher the strain rate the higher the stiffness, higher resistance to further strain application
- loading as well as unloading are superimposed by stress relaxation phenomena if the tests runs strain controlled.
- Loading and unloading are superimposed by creep phenomena if the tests runs load controlled

These phenomena will result in a hysteresis loop showing deviations from pure elasticity.

After unloading. If the load is removed completely a **remaining strain** is observed.

The viscoelastic behavior may be modelled as a network of spring and dashpot elements.

A spring should be ideallly elastic, in case of linearity the force is proportional to the elongation

$F=k*\Delta l$ or $\sigma=E*\varepsilon$

For a dashpot the force is proportional to the strain rate

$F=\kappa*dl/dt$ or $\sigma=\eta*d\varepsilon/dt$

With reference to the viscoelastic behavior of the tissue components

Elastin is reported to have almost linear elastic behavior in a large range of strains (up to 100%).

Collagen is viscoelastic

The *proteoglycan matrix* is viscous (lubricant)

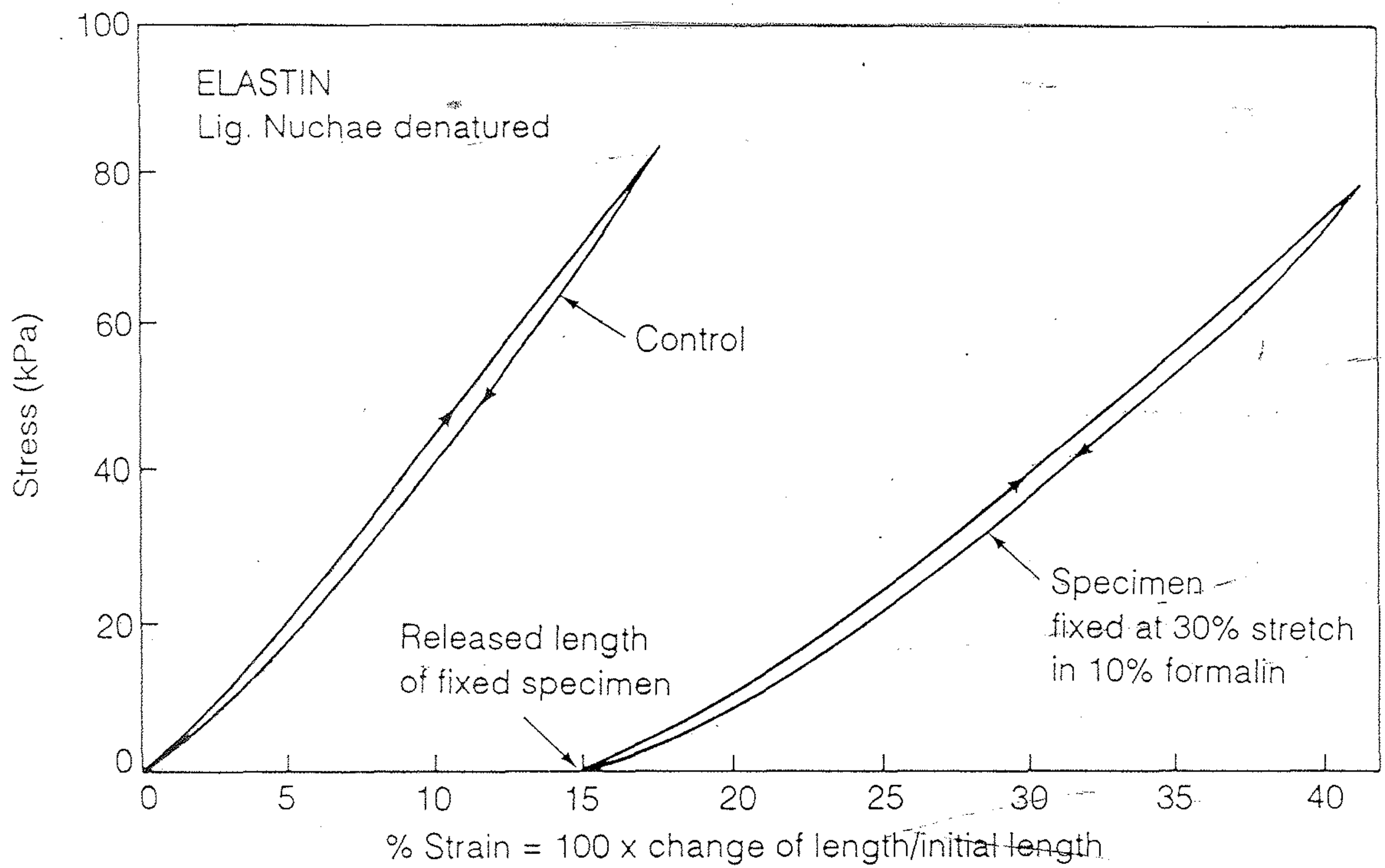
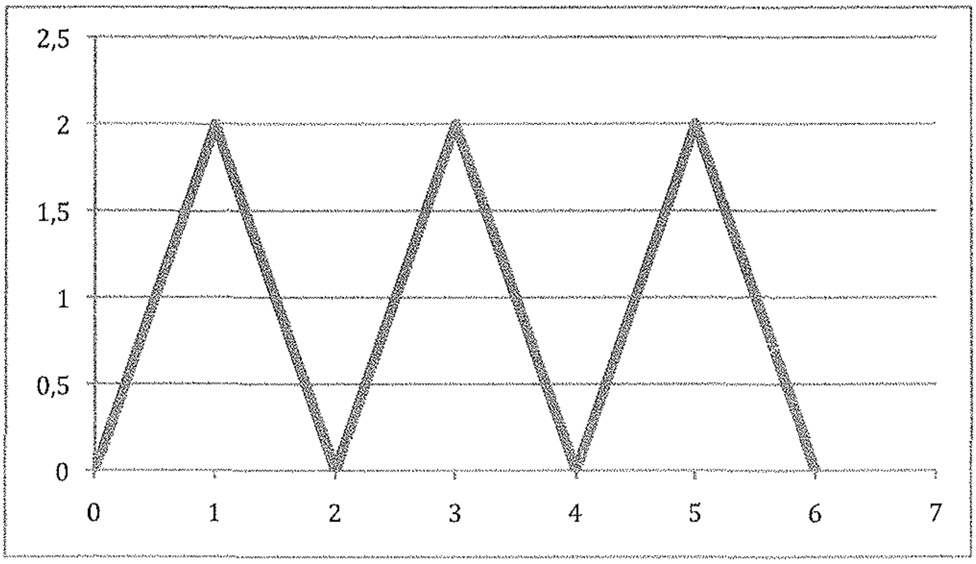


Figure 7.2:2 The stress–strain curve of a specimen of elastin that was first stretched 30% and then soaked in 10% formalin for two weeks. On releasing the stretch, the specimen shortened, but 15% of stretch remained. Subsequent loading produced the stress–strain curves shown on the right-hand side. These curves may be compared with the "control." The arrows on the curves show the direction of loading (increasing strain) or unloading (decreasing strain). From Fung and Sobin (1981). Reproduced by permission of ASME.

Further schemes of experiments to characterize the viscoelastic behavior of bio-tissues

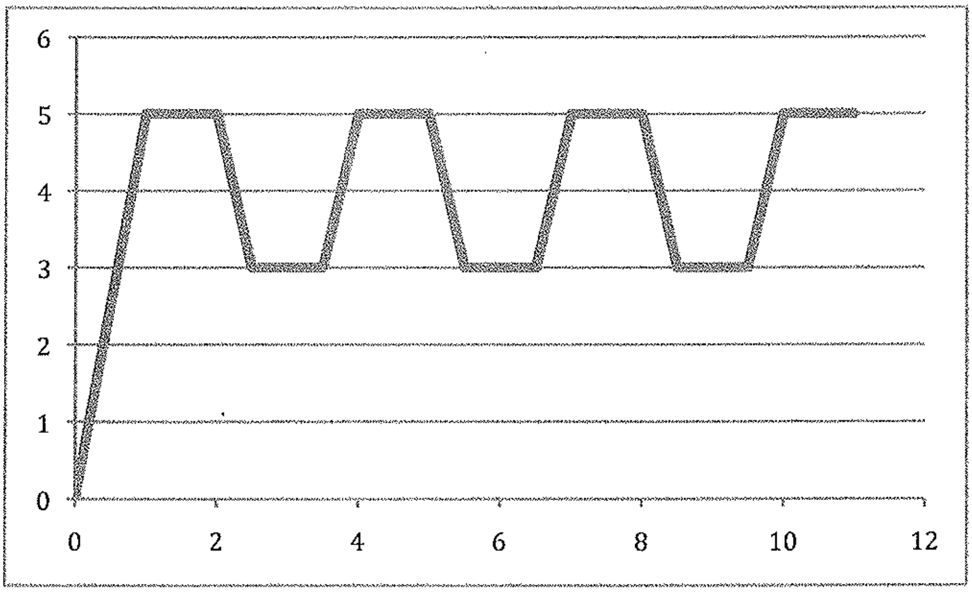
Cycles of repeated loading and unloading (**preconditioning**). The experiments are done until a stionary state is achieved. Thus the loading as well as the unloading branches do not alter any more from the n.th to the follwing (n+1).th cycle



X-axis = time

Y-axis = strain

Smilar experimental procedures by rectangular shaped straining or sinusoidal straining



X-axis = time

Y-axis = strain

The stress-strain relationships found by **loading and the unloading remain different** when the stationary state is achieved. Thus constitutive equations are different for loading and unloading even after preconditioning.

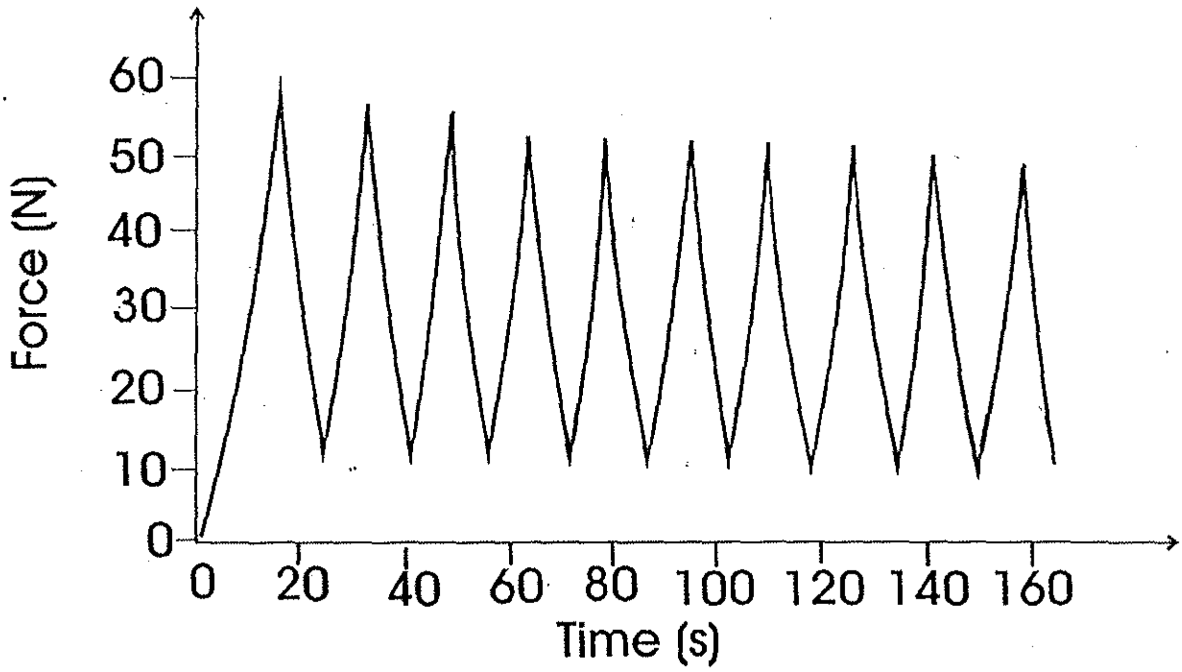


Figure 7 -Shown is a typical ligament response to cyclic tensile loading and unloading. Peak loads decrease with each cycle indicating ligament softening.

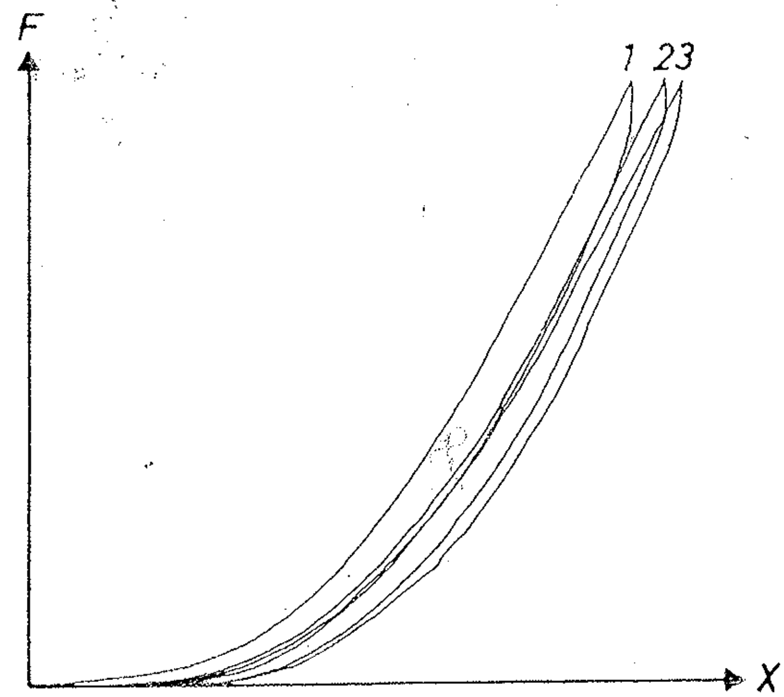


Fig. 13. Load-deformation diagrams (cf. Fig. 14) for three consecutive cycles of loading and unloading of the same specimen.

The following diagrams show **stress relaxation tests** superposed by **strain pulses**.

The first plot shows to subsequent stress relaxation tests after loading steps.

The second plot shows a relaxation test after an unloading step. In this case the load increases with time

The third plot shows the influence of changes of the frequency of the strain pulses. After the frequency has been altered the specimen has to be preconditioned anew to achieve a stationary state.

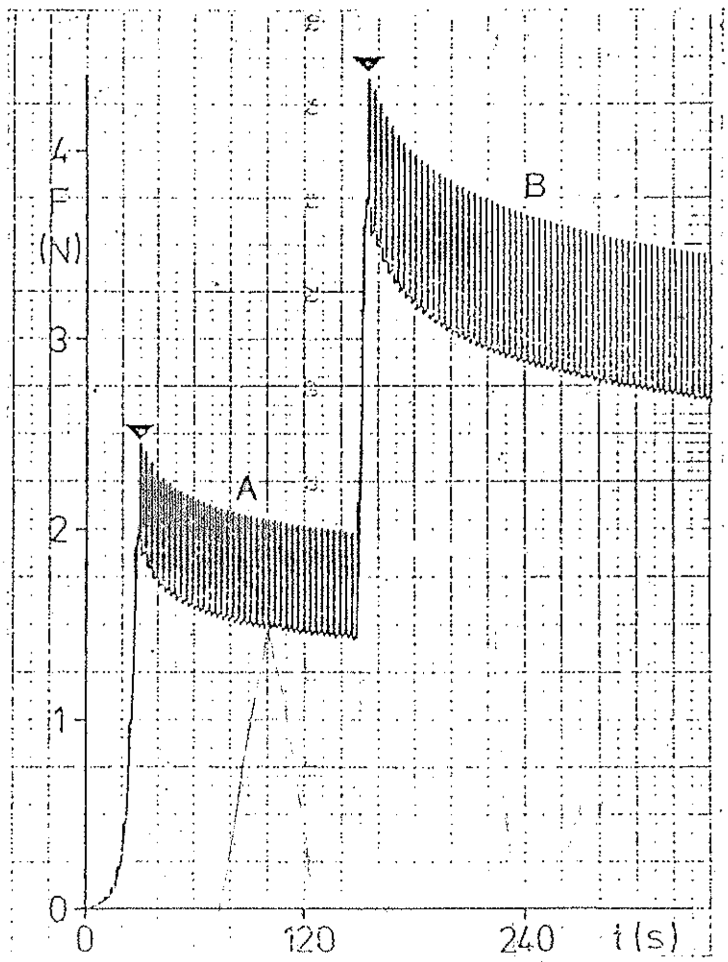


Abb. 5. An einem Längsspaltteil einer Sehne vom Rind wurden impulsmechanische Spannungs-Relaxations-Versuche ausgeführt. Lineare Längenzunahmegeschwindigkeit $v_{\Delta l} = 7{,}5$ mm; 0,9%ige NaCl-Lösung; Temperatur $\vartheta = 29$ °C; thermische und mechanische Vorgeschichte: keine Schädigung, viskoelastischer Gleichgewichtszustand; Längenimpulse $H = 0{,}17$ mm. Während der Impulsrelaxation A und B wurde der lineare Streckungsvorgang jeweils konstant gehalten.

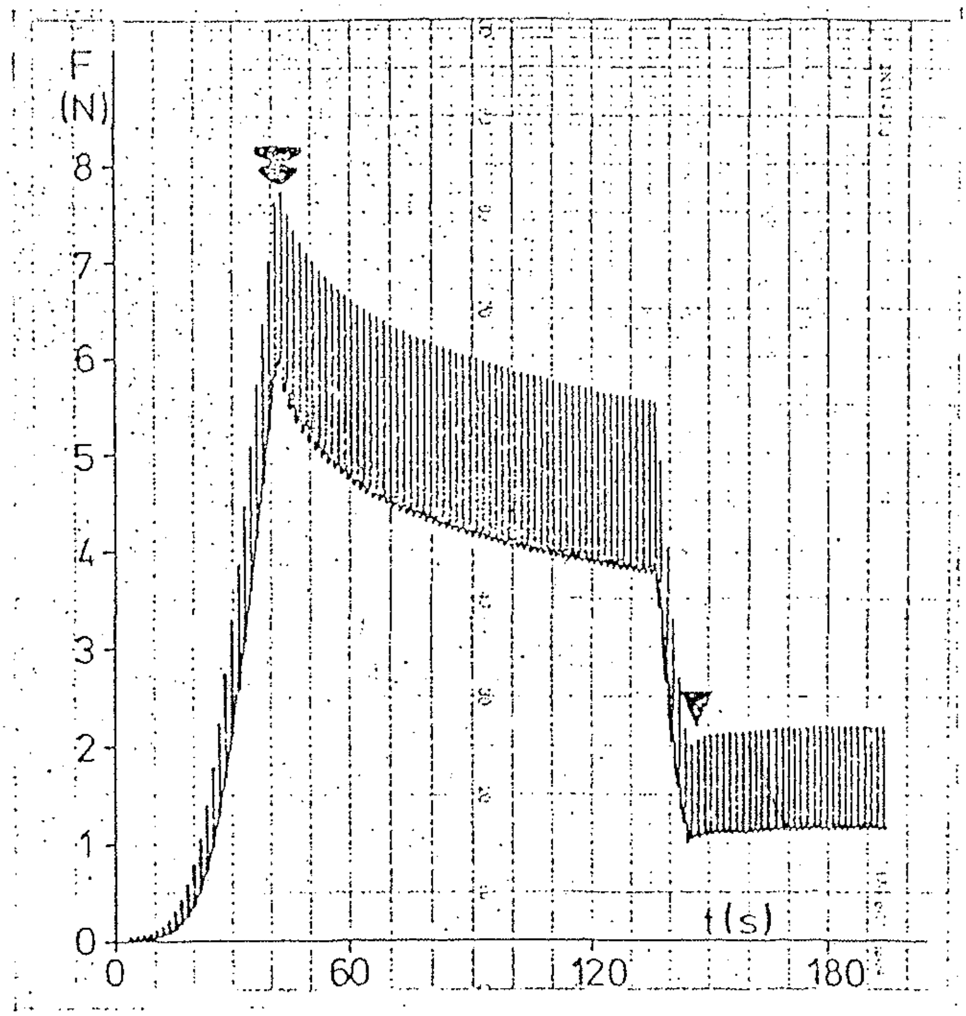


Abb. 6

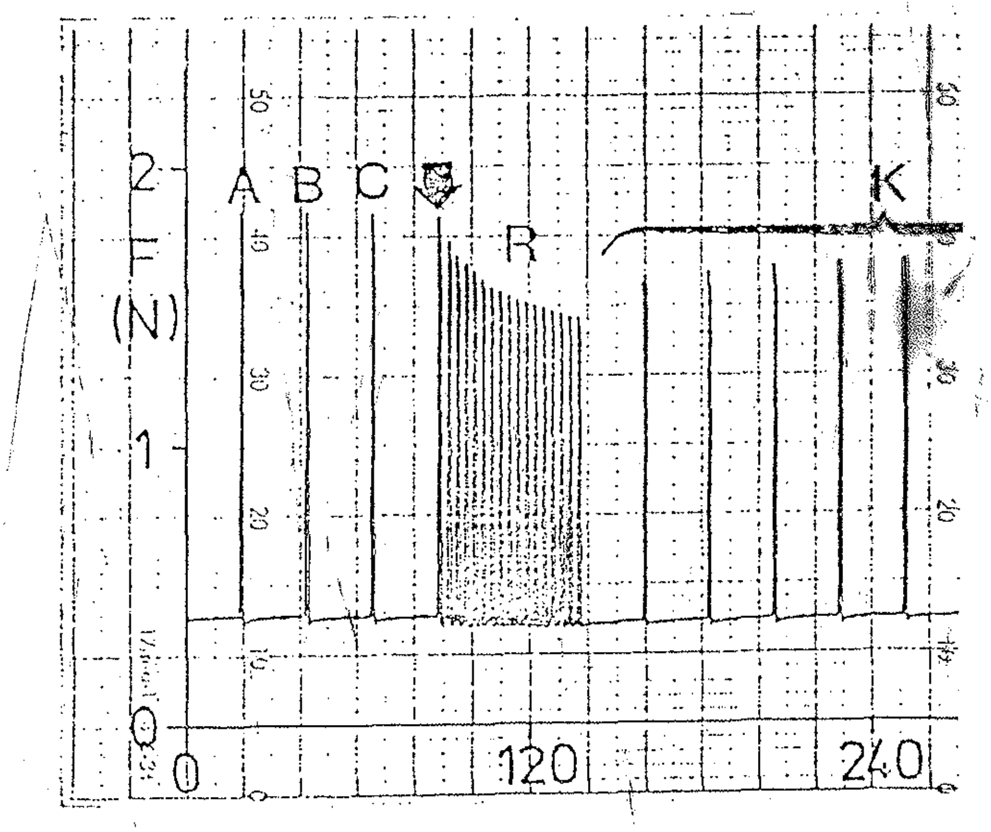


Abb. 7

Tendons and ligaments have similar structures

Tendon connects *muscle and bone*

Ligament. *bone-bone* connection

The structure is characterized by almost parallel fiber bundles of **collagen.** The collagen bundles are cross linked to **elastin**

The amount of *crosslinking collagen-collagen* depends on

- age of the individuum
- enhanced glycation (diabetes)

Experiments with non enzymatic glycation and inhibiting agents are amply reported->lit.

The **proteoglycan matrix** is reported to act as lubricant for the gliding mechanism when the tissue is strained

Tendons transfer the force exerted by muscles to the skeleton

(Short) tendons with a large cross-sectional area for transmitting high forces (e.g. quadrizeps)

(Long) tendons with lower cross-sectional area for lower loads and to enable complex movements (the fingers)

Influence of immobilization and exercise

Incresed or decreased physical activities affects profoundly the structural and mechanical properties of ligaments

- it is reported to take 300-500 days for changes in *tendon structure* when performing exercise *larger* for structural changes observed in *bone*
- The strength is decreased dramatically due to immobilization

Exercise and Disuse

Professional and recreational athletes experience periods of increased and decreased physical activity. These cycles are, in part, dictated by competition schedules and injuries. Alterations in activity levels can have profound effects on the structural and mechanical properties of ligaments.

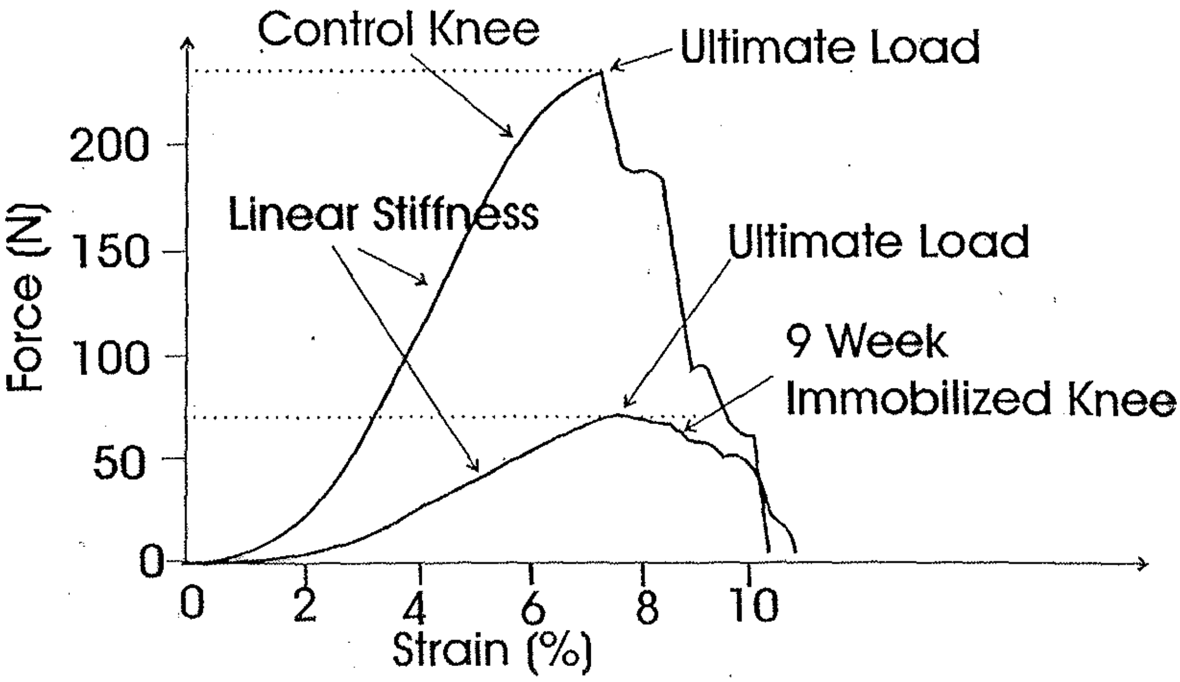


Figure 9 - Effects of 9 weeks of joint immobilization on the rabbit medial collateral ligament. Ultimate failure load decreases along with tissue stiffness and the energy absorbed prior to failure.

Structural and mechanical properties are affected by aging

- Distinction is made between maturation and aging
- The **cross-sectional area** of tendons **increases during maturation** then remains constant-> diagram
- The **water content** and content of **elastin is decreased** with age. The flexibility of tissues is decreased
- The bone-ligament compound is weak compared to the ligament itself for young individua. In the old age it´s vice versa, obviously by different rates of aging
- The elastin is coarsened in the old age
- The crosslinking of collagen is enhanced

The effect of structural changes has consequences for the stress-strain behavior as shown for rat tail tendons and rat skin samples -> following diagrams

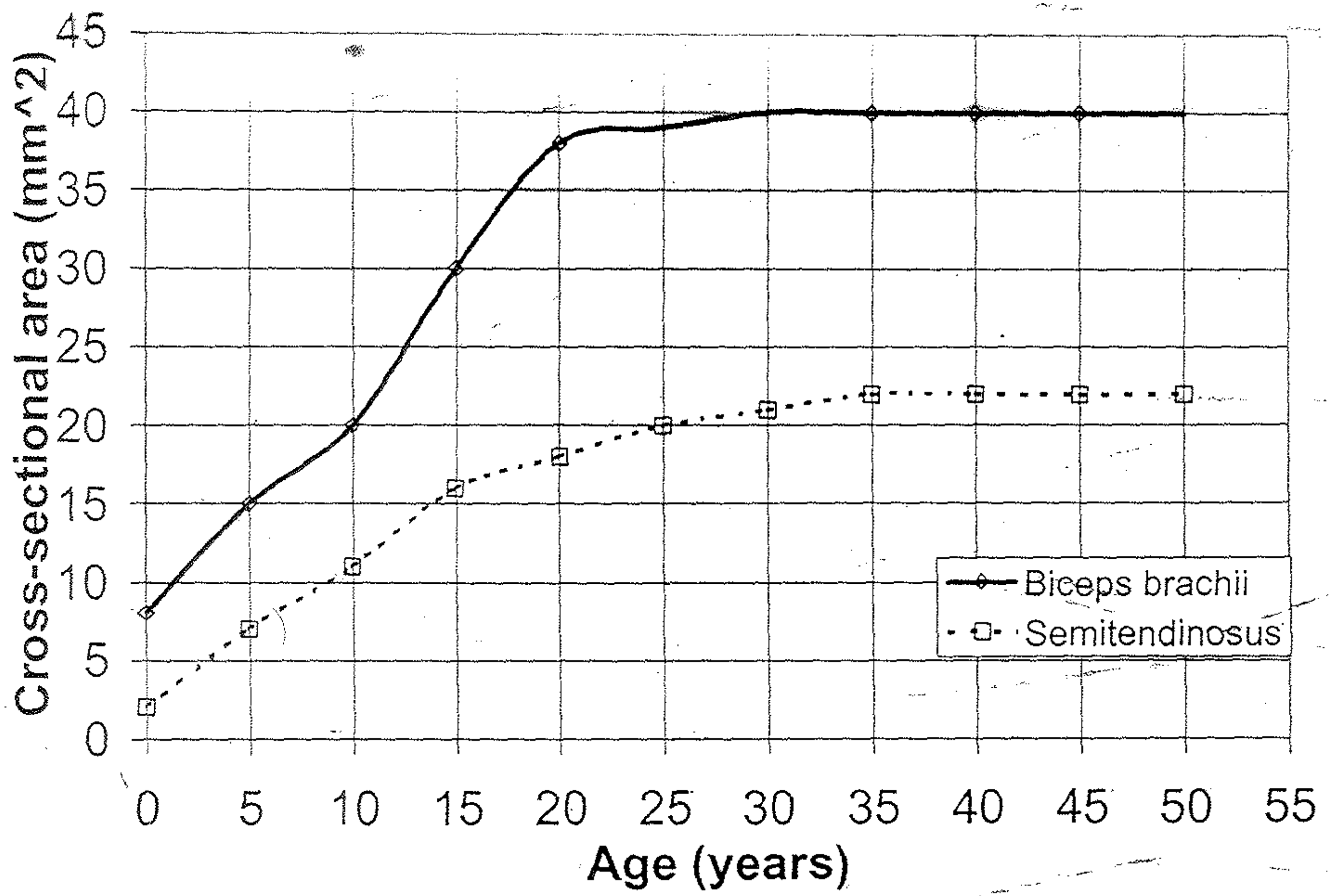


Figure 2: Tendon cross-sectional area increases to adulthood ~~and decreases~~ in later life.

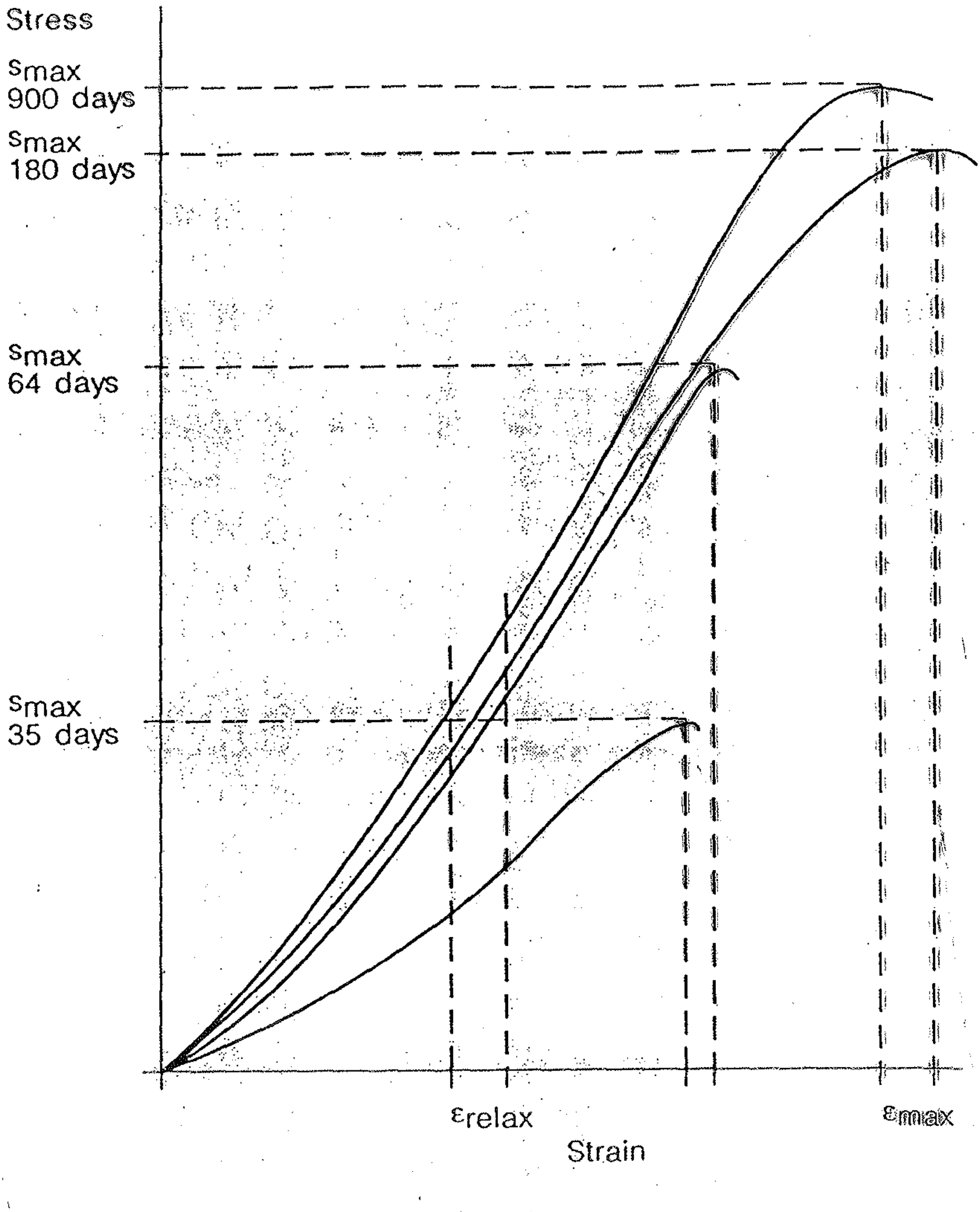


Fig. 1. Stress-strain relationship of RTTs from animals with different ages.

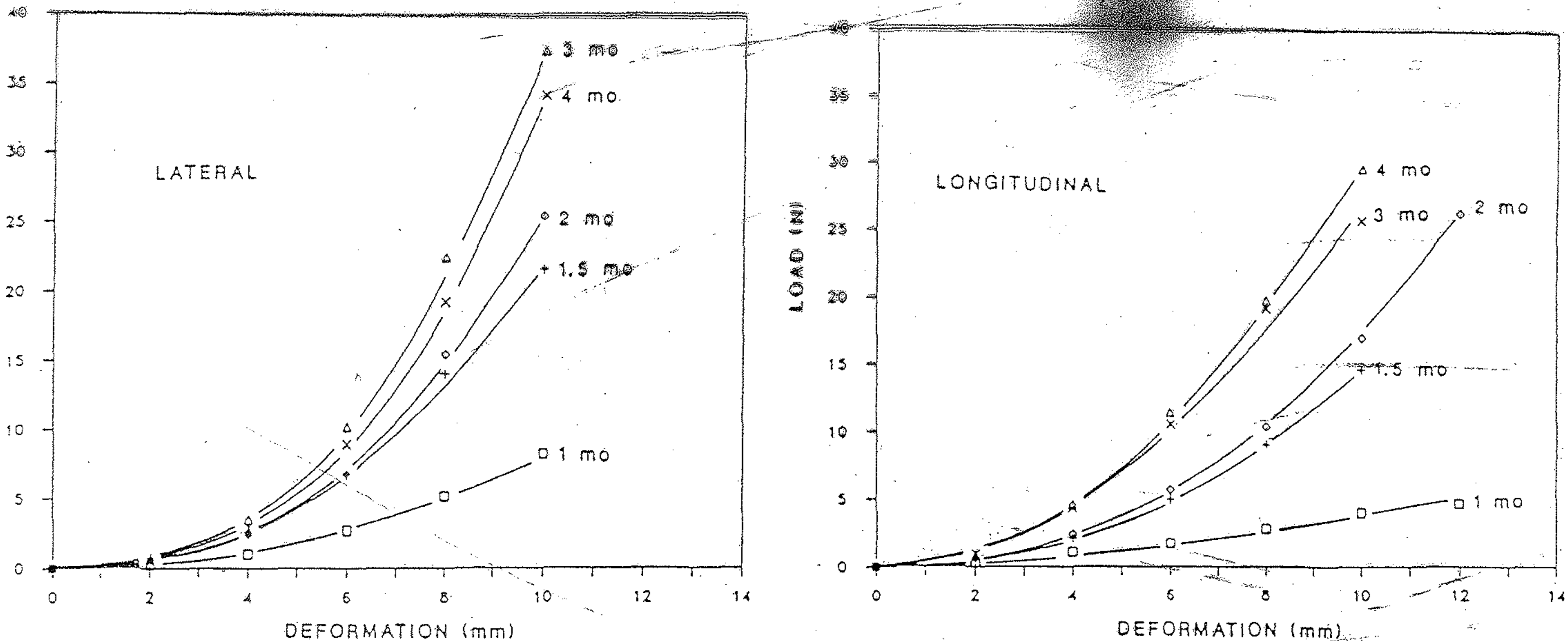


Fig. 2. Average tensile response of skin specimens oriented laterally and longitudinally. Averages were taken at deformation increments of 2 mm.

Also thermal contraction experiments were performed to characterize the influence of age to tissue properties. Temperature is raised at a constant rate using differential calorimetry (DSC) and the heat flow to the specimen is plotted versus temperature compared to a reference specimens which does not undergo a phase transition within the temperature range of interest. The collagen is transformed from a regular arrangement of fibers to an irregular one. This is a phase transition which alters the heat flow showing a peak at the transition temperature. This temperature increases with age of the rat tail tendon samples -> figure

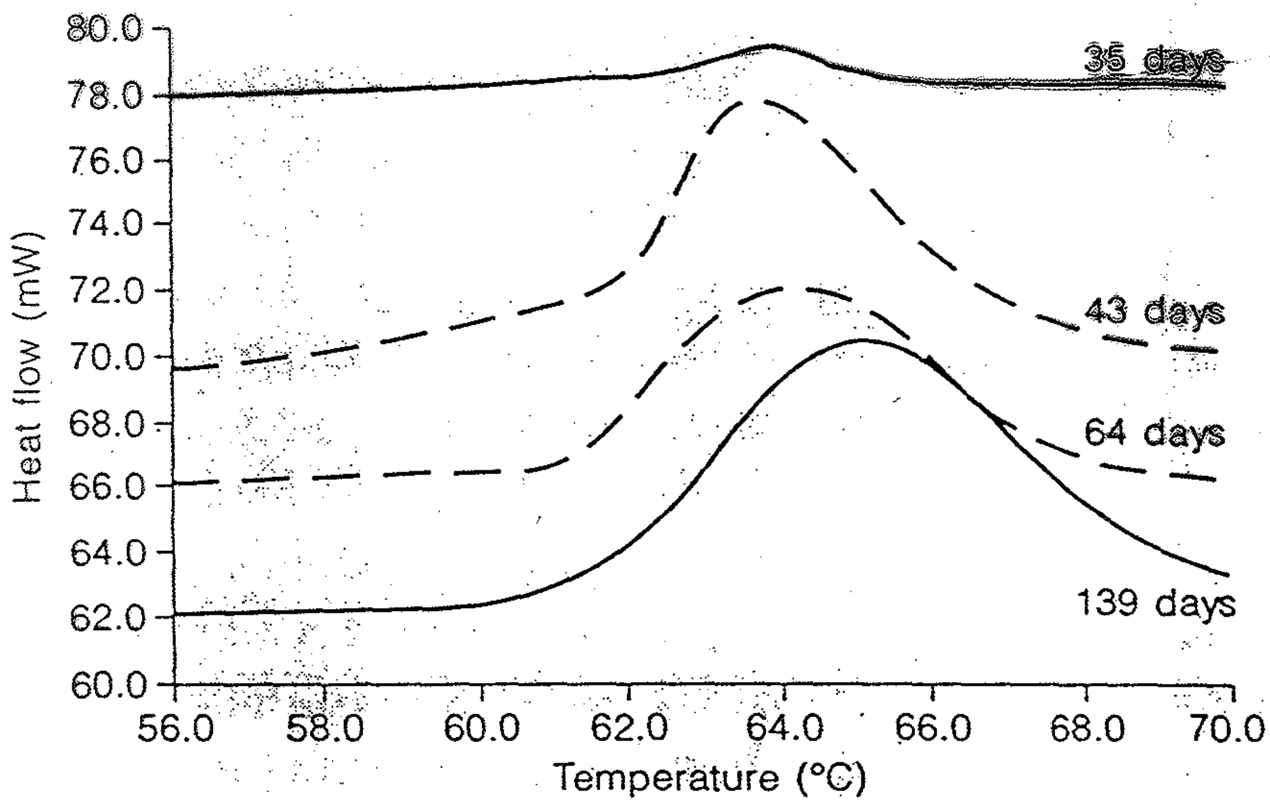
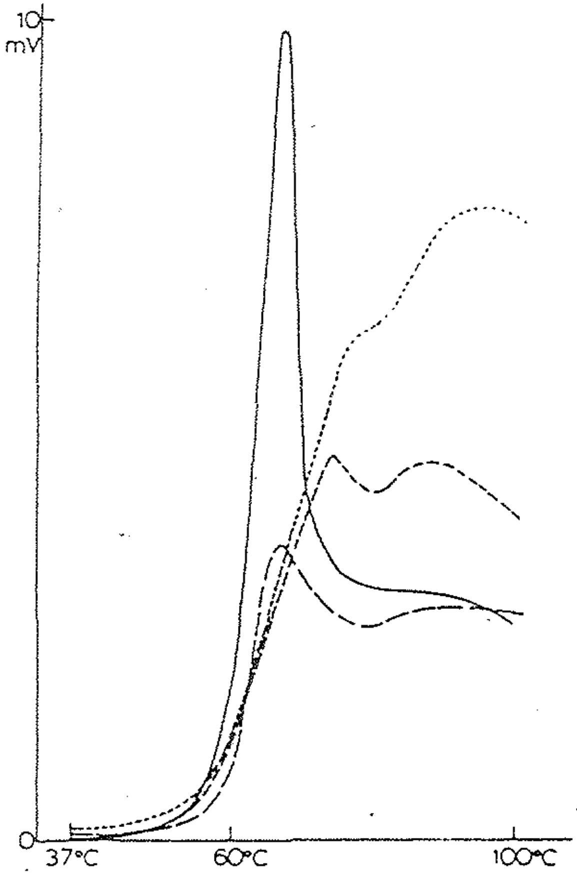


Fig. 2. Increase of collagen denaturation temperature as measured by DSC (heat flow vs. temperature).

It can be seen that the maximum tension was reached at a much higher temperature in skin of new born ~~animals~~ (95° C) than in adult animals (67° C). In



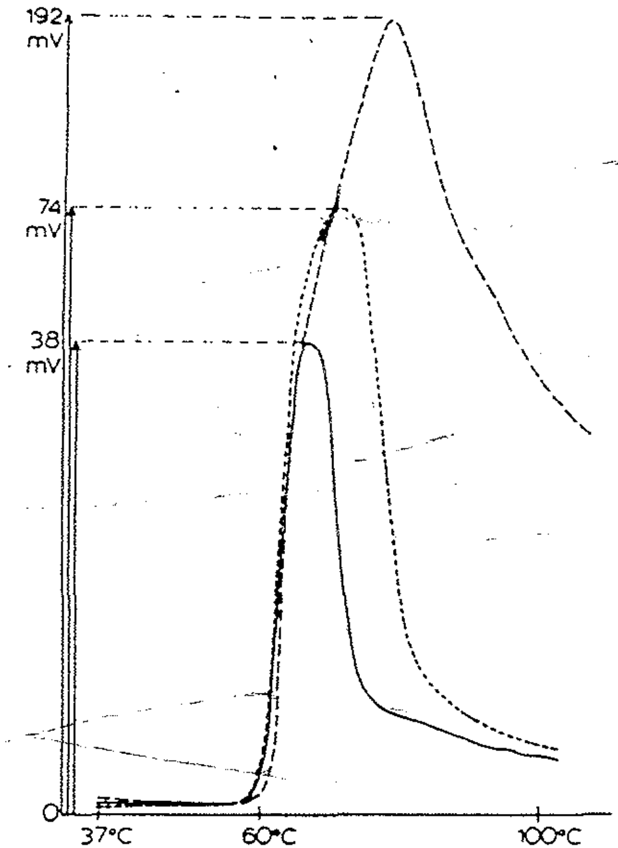


Fig. 2. Modification of isometric tension between 37°C and 100°C in rat skin from birth to 1 month: - - - - - -, new born rat; – – – –, 1 week; —— —— ——, 3 weeks; ————, 1 month old rat.

Fig. 3. Modification of isometric tension between 37°C and 100°C in rat skin from 2 to 18 months. ————, 2 months; - - - - - -, 5 months; – – – –, 18 months.

Now to the mode of failure as observed with preparation bone-ligament-> diagram

- at **high strain** rates the ligament itself ruptures (**tearing**)
- at **low strain** rates pieces of the bone break away with intact ligament (**bony avulsion**)
- The bone-ligament interface fails only in few cases obviously due to perfect compound bone-ligament

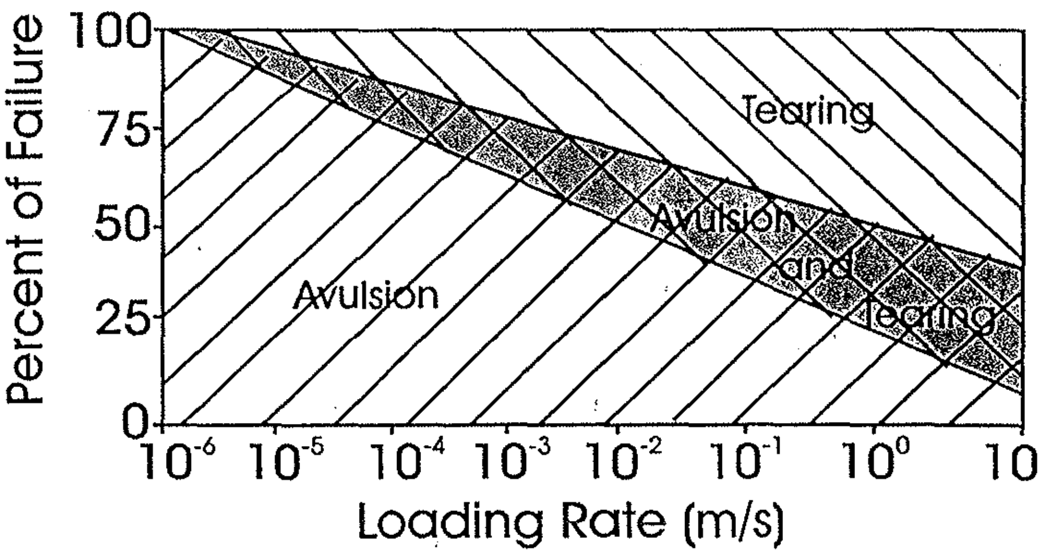


Figure 6 -Experimental results are presented indicating the probability of specific failure modes of bone-ligament-bone units as a function of loading rate. At slower loading rates bony avulsion failures have the greatest probability of occurring. At fast loading rates mid-substance failures are to be expected. (Modified from Crowninshield, R.D. and Pope, M.H., The strength and failure characteristics of rat medial collateral ligaments. Journal of Trauma, 16(2), 99-105, 1976)