3. Biomechanics of Materials

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Introduction

The materials orthopaedic surgeons are mainly concerned with: muscle, tendon, ligament, bone, and cartilage and indeed the materials which, by historical accident, dentists are concerned with—enamel and dentine—can be divided functionally into two classes, the active materials, and the passive ones. Muscle is active; it can transform chemical energy into force and work. The rest are passive. The extent to which they enable the body to live in the hostile world in which it finds itself is almost entirely the result of their passive mechanical properties. Of course, most are living tissues, and are capable of modifying themselves over time, in response to the mechanical, chemical and biological environment. But, usually, when the environment changes in a matter of seconds, or milliseconds, the mechanical properties of the tissue are of paramount importance. In this review I shall not deal with muscle or dental tissues, but confine myself to bone, cartilage, ligament and tendon. I shall assume that the reader is familiar with the broad outlines of the subjects discussed by Tony Unsworth in his review of Biomechanics. Books by myself[1] and Silver[2] provide reasonably simple introductions to the biomechanics of bone and of soft tissues respectively.

The Important Mechanical Properties

Stiffness

In a well-behaved material, Young's modulus of elasticity is the most important measure of stiffness. However, quite often Young's modulus does not give all the information needed. Figure 1 shows typical tension load-deformation traces for bone and tendon. The trace for tendon is curved over much of its length, and therefore Young's modulus, which is defined as stress/strain, is not the same at all strains; instead one can talk about the 'tangent modulus' which is Δstress/Δstrain measured at the point on the trace that is of interest. Such J-shaped traces are characteristic of compliant materials, like tendon, ligament, cartilage and skin.

Compliant materials also tend to show strain-rate sensitivity; that is, their measured mechanical properties vary according to how quickly they are loaded. Tendon, if loaded quickly, is stiffer than when loaded slowly. Therefore, not only is there no unique Young's modulus for tendon, but even the tangent modulus at any particular stress will vary according to the strain rate.

Finally, in order to understand the mechanical properties of a material, we often need to know about another type of stiffness: the shear modulus, or modulus of rigidity. The shear modulus is a measure of how resistant the material is to being kinked or twisted. The highest values of the tangent modulus of tendon (about 1.5 GPa) are about one tenth the value for adult cortical bone (15 GPa). However, tendons can easily be tied into a knot, with a very small radius of curvature, without being damaged. Bone, on the other hand, is stiff in bending. The shear modulus of tendon is extremely low, far less than that of bone. The difference is like that between a steel rod and a wire rope of the same diameter. The rope will be somewhat less stiff in extension than the rod, but can be coiled in a way quite impossible for the rod.

Strength

Strength, also, is not quite as simple a concept as might appear. The stress-strain traces for tensile

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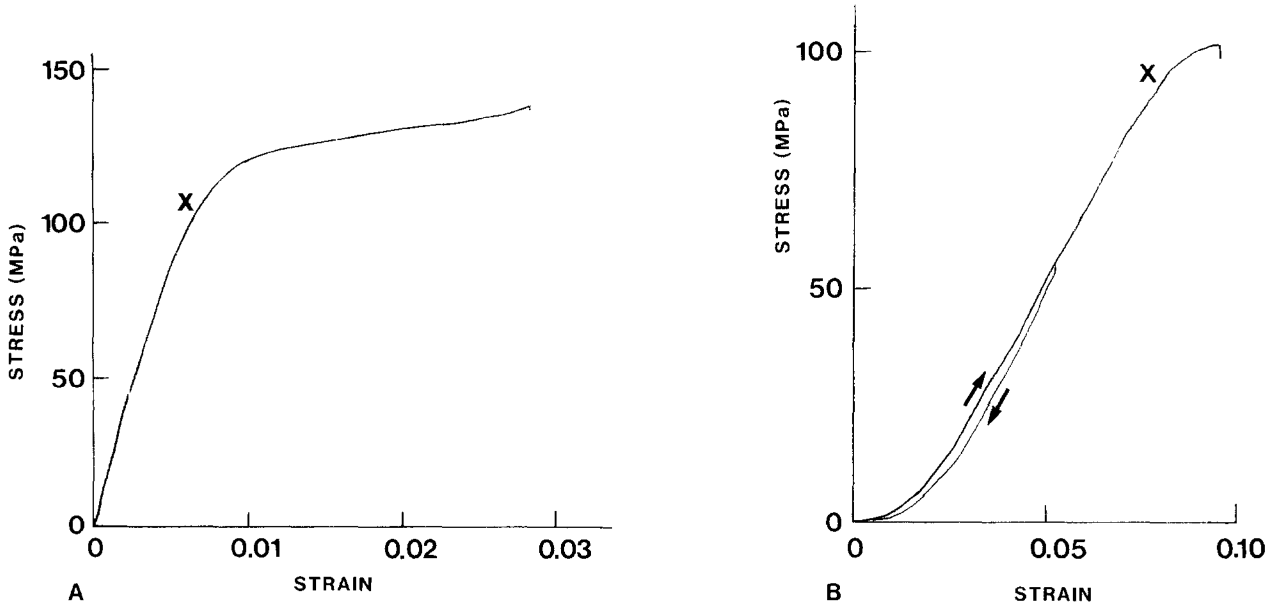


Fig. 1—Stress-strain traces of compact bone (A) and tendon (B), loaded in tension. Note that the scales of the axes are different in the two figures. The behaviour of tendon on unloading is also shown (arrows). The fact that the unloading trace is so close to the loading trace shows that tendon pays back nearly all the energy put into it.

loading in Figure 1, for bone and tendon, both have a highest point, and the strength of the materials could, perhaps be said to be the stress at this point. However, somewhere near the point marked X on each trace the material has started to undergo irreversible damage. This is called the yield point. Even more striking is the stress-strain trace for compressive loading of cancellous bone (Fig. 2). The trace rises, drops slightly, and then shows a considerable increase in strain, finally the stress begins to rise rapidly. What is happening here is that at about point X some individual trabeculae start to collapse, and the long flattish region marks the progressive spread of this collapse through the specimen. The final rise of stress comes when the specimen is so deformed that it becomes compacted. When this stage is reached, the material is unlike what is ever found in the body. It is as if previously we had been deforming the branches of a tree, and are now trying to deform a stack of logs.

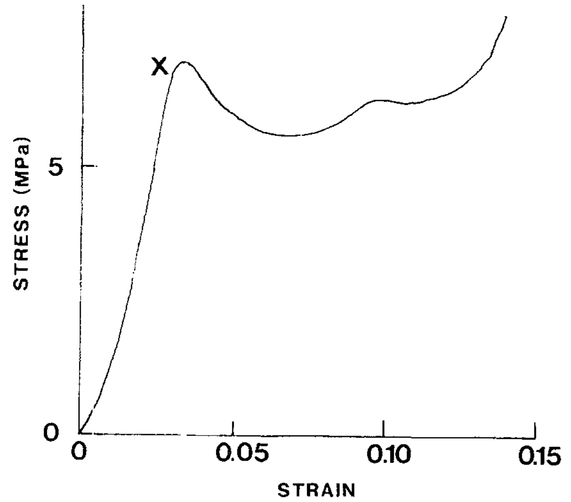


Fig. 2—Stress-strain trace of a block of cancellous bone, loaded in compression.

Are we interested in the maximum stress a material can stand, or the stress it can withstand without being damaged at all? It depends, of course, on the circumstances. In treating people, we are usually concerned with loads that can be borne again and again. Here, it is vital that no damage occurs, or it will gradually accumulate with time and eventually break the tissue. Occasionally in real life the question is simply whether, under some extreme loading, a bone will break or not, or a tendon will snap or not. If it does not, then it may be repaired later by remodelling, even if it has been damaged. If it does break, the case is much more serious.

Fatigue strength

Often a material can be loaded repetitively to a stress considerably less than that at which it appears to be damaged and yet, after many repetitions of this load, the material begins to show signs of damage, and may break. This kind of damage is called fatigue damage. In living tissues, of course, the spread of fatigue damage may be prevented by biological repair processes. For some reason I have never been able to fathom, orthopaedic surgeons call such a fracture in bone a stress fracture, as if this distinguishes it from other kinds of fracture.

Elastic energy absorption

In ordinary locomotion the tissues of the body are continually being given mechanical energy, and it is sometimes useful to know how much energy a material can absorb without being damaged in any way. The area under the stress-strain trace of a material tells how much energy it absorbs up to any particular point, and the area under the unloading trace shows how much of this energy it pays back. The difference between the two traces shows the amount of energy that is lost as heat etc. (Fig. 1B). Usually, as we shall see, it is best if little energy is lost.

Toughness

The 'toughness' of a material is a measure of its ability to absorb damage, to absorb energy without breaking, and to resist the spread of any cracks that may exist in it. It is a difficult concept to define clearly, and a whole sub-branch of the study of the strength of materials, called fracture mechanics, is devoted to determining the toughness of materials. Materials that may appear strong if they are prepared with smooth surfaces, may be quite weak if they possess tiny cracks. Glass is a material like this. An indication of the toughness of a material is given by its stress/strain trace. A material that breaks as soon as its trace begins to bend over, at the yield point, is said to be brittle, and will probably not be very tough. A material that has a long post-yield region is likely to be tough. The reason for this is that the area under the stress/strain trace is a measure of the energy absorbed by the material. Therefore, a long post-yield region implies that much energy is absorbed after the material has yielded, even though after the yield point the material is damaged. This is shown particularly clearly by bone (Fig. 1) although, in fact, bone does not absorb a great deal of energy overall, and is not very tough. It is, however, much tougher than it would be if it broke as soon as it yielded.

General Comparisons

The mechanical properties of various skeletal materials are shown in the Table. Although the absolute values of the different properties are given, the relative values are of more interest. Compact bone has a higher Young's modulus than the other tissues, it is stronger in tension, and much stronger in compression. On the other hand, although we do not know much formally about the toughness of tendon, it is certainly much much better at absorbing energy without breaking than is bone. Cartilage does not seem to do very well on any count except that it is good at absorbing energy but it has, of course, other virtues. Cancellous bone has an enormous range of properties.

Bone

Compact bone

Compact bone has as its main function to be a stiff material. Although it is stronger than tendon in tension, it is not very much superior when it is considered on a per weight basis. Furthermore, compared with tendon it is not very tough. However, in its use as a lever to push against the environment, as a rigid box to protect the brain, and in all the other functions that require something to be stiff in bending, it is unrivalled. The stiffness of bone is produced by the deposition of apatite in and around the fibrils of the collagen matrix. In different bones the amount of mineral varies, and in general the more mineral it has, the stiffer it is but also the less tough it is. This is taken to one extreme in the otic bones, which have a high mineral content, and are very stiff and brittle. However, their brittleness does not matter, because they are fairly well protected inside the skull and are not exposed to large loads. Their stiffness is valuable for reasons to do with hearing.

The fact that bone in general is not very tough means that it is rather 'notch sensitive'. If one makes a hole in a material and then loads it, the forces are concentrated in the vicinity of the hole, and so the stress is locally higher than in the rest of the material. A tough material can diminish the effects of this stress concentration, but a fairly brittle material like bone cannot so well, and as a result the load at which a bone with a hole in it will fail is much less than in a similar bone without a hole. For this reason one should be aware that making holes in bone, even if one fills the holes with screws, has mechanically very bad effects on the bone.

Highly mineralised bone is stiff and brittle. Conversely, lightly mineralised bone is less stiff but tougher. This is seen very clearly in changes that take place during life. The bones of young children are less mineralised than those of adults, and have a lower Young's modulus, of the order of 10 GPa or less, compared with 15 to 20 GPa. However, they are tougher. The impact energy absorption (a convenient if not very rigorous measure of toughness) is about twice as great in 5-year-old children as in adults.[3] This shows itself clinically in greenstick fractures.

Although we know the tensile and compressive failure properties of bone, in the real world bone

Table—The mechancial properties of various tissues. The earbone is the tympanic bulla of a whale, but the amount of mineralisation of human earbone is similar to that of the whale, and it is likely that their mechancial properties will also be similar

Material	Young's modulus (GPa)	Compressive strength (MPa)	Tensile strength (MPa)	Ultimate strain (%)	Elastic energy absorption ($J\,m^{-3}$)
Costal cartilage	0.1	—	20	8	5×10^5
Synovial cartilage	0.012	—	20	50	5.6×10^7
Tendon	2–1	—	100	10	2×10^6
Compact bone	15	250	170	2	3×10^5
Ear bone	30	?	10	0.2	1.7×10^3
Cancellous bone	4–0.5	50–1	20–1	10	5×10^4

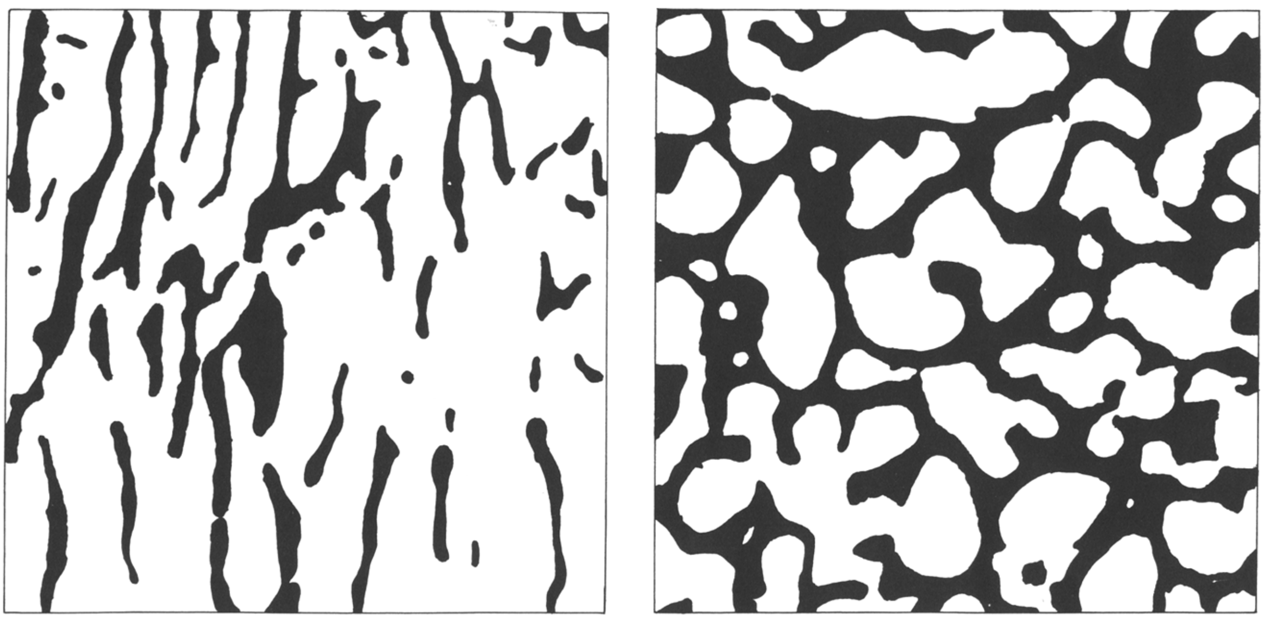
nearly always break in tension, except when it forms the thin shell round cancellous bone. This is because, as Tony Unsworth explained, bending produces similar tension and compression stresses on opposite sides of bones and, because bone is about 50% stronger in compression than in tension, it is the tension side that breaks. Long bones, in particular, tend to break because they have been bent, though torsion may occasionally be important.

Cancellous bone

Cancellous bone is always weaker and less stiff than compact bone. It is often thought that this is compensated for by the fact that it is lighter. However, it is easy to show that, if a bone of minimum mass is required to carry out some particular function, then it is always better to have it made of compact rather than cancellous bone. This becomes even more the case if the mass of the marrow is taken into account. Nevertheless, a combination of compact bone and cancellous bone is often lighter than a structure made of one type alone would be. This is true of the diploë of the skull, and also of short bones like wrist and ankle bones, and the vertebral bodies. Unfortunately, the argument becomes more convoluted in the case of the cancellous bone under the ends of long bones, and it is possible that here the greater flexibility of cancellous bone, when loaded in compression, may help to spare the articular cartilage from having to absorb too much energy when the joint is loaded.[1]

Several features of cancellous bone effect its mechancial properties. Most important is its 'apparent density' (the mass of bone in a given anatomical volume). This, more or less, is what is measured by X-Rays etc. The strength and stiffness are not proportional to the apparent density, but roughly to its square, so a block of cancellous bone of three times the apparent density of another will be nine times as stiff and strong. Another feature of some importance is the fabric or architecture of the bone. How the trabeculae are arranged in space will have an effect quite separate from how thick they are. For instance, if the trabeculae are drawn out in one direction, giving the bone a definite preferred orientation, it will be stiffer in that direction, and less stiff in directions at right angles to it. A more randomly or evenly arranged set of trabeculae will tend to be uniform in their mechanical properties, whatever the direction of loading (Fig. 3). A final variable that has some effect on the mechanical properties is the mineral content of the bone material itself. This is the most important variable in compact bone, but in cancellous bone it is almost overwhelmed by the effects of variation in density and fabric.

Putting it in statistical terms, we find that if we take many samples of young human cancellous bone, about 65% of the variation in mechanical properties is produced by variation in apparent density, about 25% by variation in fabric, and 2% by variation in mineral content. The remaining 8% or so is accounted for by experimental error and features of cancellous bone that have not yet been discovered. As yet we do not know whether old cancellous bone is qualitatively different from younger bone, or whether it can be explained in the same terms. Of course, for any particular site, older cancellous bone usually has a lower apparent density, spectacularly so in cases of



A B

Fig. 3—Faces of two cubes of human cancellous bone to indicate the effect of apparent desity and fabric on Young's modulus. Bone is shown black. Cube A obviously has a lower apparent density than cube B, and is more highly oriented in the top to bottom direction. Cube A had a Young's modulus of 400 MPa when loaded top to bottom, and 79 MPa when loaded from side to side, a ratio of 5:1. Cube B had a Young's modulus of 1105 MPa when loaded top to bottom, and 1540 MPa when loaded from side to side, a ratio of 1.4:1. The faces show only the bone that lies at the surface; in reality all the apparently unconnected bits of bone would be connected somewhere in the third dimension. Note that only one of the six faces of each cube is shown, but they give a good idea of the general arrangement of the bone in the cube, and of the difference in apparent density.

osteoporosis, with a concomitant reduction in stiffness and strength.

Tendon and Ligament

For a number of reasons it is much more difficult to be sure about the mechanical properties of tendon and ligament than about those of bone. Their load-deformation traces are curved, and so we have to use a tangent modulus; they are more strain-rate sensitive; they are probably much more sensitive to temperature effects than is bone; unless one can test ligaments, attached at both ends to bone, there are problems in gripping, and measuring deformation; finally, there is the great difficulty, found in all soft tissues, of the effects of 'conditioning'; when a piece of soft tissue is first tested, it often shows mechanical properties that differ greatly from those shown on subsequent tests.[4]

Nevertheless, the properties of tendon shown in Figure 1B, and in the Table, are probably not too far from the truth. There seem to be no consistent differences between the properties of tendon and ligaments. The J-shape of the stress-strain trace is, as has been mentioned above, often found in soft tissues. In tendon and ligament it seems to be caused by the straightening out of the initial crimping present in the collagen fibrils. Although it would be possible to measure the compressive strength of tendon or ligament, it would not make much sense to do so, because they are never functionally loaded in compression.

Energy absorption and safety factors in tendon

Figure 1B shows that the unloading trace for tendon is quite close to the loading trace. This means that rather little of the elastic energy stored in the tendon on stretching is lost as heat when it relaxes. This is important for two reasons. One is that, if much of the strain energy in the tendon were lost as heat every time it was unloaded, its temperature would rise, possibly dangerously, during rapid locomotion. The other, more important, reason is that if the elastic energy is returned on unloading, tendons can be used as efficient energy stores, acting like springs. Despite being weaker, tendon is able to store much more energy than bone. Bone can safely be loaded to a stress of say 100 MPa without undergoing any damage, and has a Young's modulus of 15 GPa. Tendon can be safely loaded to a stress of say 75 MPa, and has a Young's modulus of 1.5 GPa. The energy stored per unit volume is proportional to the square of the stress and inversely proportional to the Young's modulus. As a result, tendon safely stores about $5\frac{1}{2}$ times as much energy per unit volume. Tendon's advantage becomes even more marked when the difference is considered on a per mass basis, because bone is almost twice as dense as tendon. On this basis the $5\frac{1}{2}$ times advantage increases to $9\frac{1}{2}$.

In a recent and beautiful paper, Ker and others[5] show that there are two functional types of tendon. One, represented in man by the Achilles tendon, is stressed in vigorous locomotion to a high proportion of its tensile strength, giving only a rather low safety factor (about 1.5). Other tendons, represented in man by plantaris, the deep digital flexors and the digital extensors, are stressed to a much lower maximum level, and seem to have a far higher safety factor (up to a factor of about 8) than is necessary. Ker et al show that these types of tendons have different main functions. The tendons with low safety factors are used during locomotion to store energy in one part of the stride, to release it in another. For such tendons, being loaded to a high stress (and so having a low safety factor) is efficient, because the amount of energy stored per unit mass is proportional to the square of the stress. On the other hand, a lightly stressed tendon will not stretch very much. If the main function of a tendon is to cause bone to follow closely the movement of a muscle, then it is better for the tendon to be lightly stressed (which is brought about by its having a large cross-sectional area) because any extension of the tendon will require an increased length of muscle to take up the slack. Muscle is weak compared to tendon, and so this increased length of muscle will be expensive in terms of extra weight, and of the biochemical work to be done by the muscle in shortening against a load. Alexander[6] has recently written a simple but fascinating book about the subject of energy storage in animals.

The low shear modulus of tendons and ligaments

In the introduction I mentioned that because tendons have a low shear modulus they can be tied in knots. The function of this low shear modulus is not to give Boy Scouts pleasure, but to enable tendons to turn tight corners. This is often necessary, particularly because tendons usually run close to joints, and indeed are kept close to the centre of rotation of the joint by retinacula. Tendons have to pull off the difficult trick of being inextensible along their length, so that the muscles that move them do not have to shorten needlessly, yet also having to accept small radii of curvature. They bring off this trick by consisting of fibrils of the rather inextensible protein collagen, arranged side by side but hardly connected, so that neighbouring fibrils can move easily relative to each other.

Attachment of tendons and ligaments to bone

Joining two materials of greatly differing Young's modulus is a formidable problem, because the strains each material undergo are very different, and this makes adhesion between the two materials difficult. Cooper and Misol[7] describe the arrangement found in the dog's patellar tendon, which is probably typical of collagenous insertions generally. Ordinary tendon merges into a fibrocartilage region, about 300 μm deep, in which cartilage cells lie in rows in the extracellular matrix of the tendon. The cross-sectional area of the tendon is correspondingly increased. Below this, starting at a distinct tideline, is a mineralised fibrocartilage region, about 200 μm deep. The mineralised fibrocartilage region merges imperceptibly into the rest of the bone, with no clear point where the fibres stop and the bone begins.

The biomechanical beauty of this arrangement is that the low-modulus material, tendon, is not merely attached to, but becomes a high modulus material, where the fibrocartilage becomes mineralised, in the space of a few microns. There are no problems, therefore, in bonding the collagen to the bone; there is continuity of the collagen fibrils from the tendon right into the heart of the bone, which is itself, of course, mineralised collagen. This is a trick that is almost impossible to bring off in man-made materials.

Cartilage

Whatever difficulties one may have in defining the mechanical properties of tendon, they are trivial compared with characterising cartilage. Articular cartilage is mostly water, and most of the mechanical properties of cartilage are determined by shape changes that are permitted, or not, by the way in which water under stress flows into or out of various regions. The mathematics of cartilage behaviour is exceptionally unpleasant.[8]

It is possible to say a few things of a rather simple-minded nature. If we test an unconfined block of cartilage, that is one that is not supported on its sides, it will have a very low shear modulus, orders of magnitude lower than that of bone. The tensile strength of cartilage, which is low[9] (Table), may not seem to be a relevant property, but in fact when two elastic bodies are pressed into each other, tensile stresses are developed just outside the region of contact. The compressive strength might seem to be a more important property; however there is no mechanism by which a cartilage sheet can fail in compression.

The most important feature of the mechanical properties of cartilage is that they are time-dependent. If a specimen of cartilage is loaded in tension, it will be possible to calculate the stress and the strain, and so derive Young's modulus of elasticity. However, with time, the cartilage will extend more and more, as water is redistributed around the specimen, and the long-chain molecules of the cartilage rearrange themselves. As a result, the calculated value of Young's modulus will drop. Behaviour like this in which a material shows elastic behaviour, but also a viscous, flowing behaviour, is called 'viscoelastic' (Fig. 4). An old, but clear introduction to viscoelasticity of biological materials is by Dorrington.[10] In fact viscoelastic behaviour is shown by all biological tissues, including bone, but is most marked in cartilage.

The mechanical properties of cartilage vary greatly according to the amount and arrangement of the collagen fibrils, and there is a continuum between extremely compliant articular cartilage and the stiff fibrocartilage found in the nasal septum and sternum. Broom[11] has attempted to relate the structure and biomechanics of synovial cartilage.

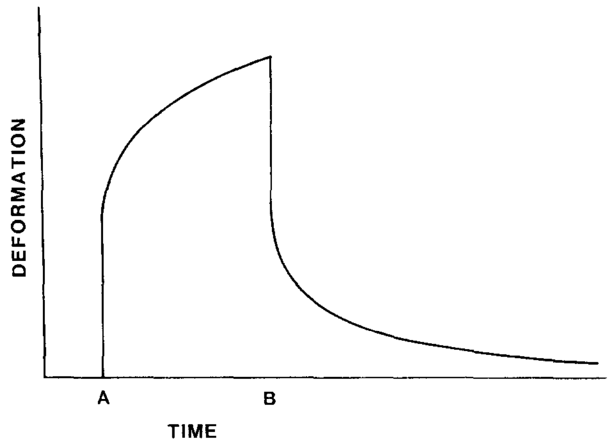


Fig. 4— Diagram showing the behaviour of a viscoelastic material such as cartilage. The ordinate is deformation, the abscissa is time. The material is loaded at time 'A', and immediately undergoes elastic deformation. The load is maintained, and viscous deformation continues, at an ever-decreasing rate. It is unloaded at time 'B'. The elastic deformation is immediately recovered, but the viscous deformation disappears at an ever-decreasing rate.

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References

1. Currey J D 1984 The mechanical adaptations of bones. Princeton University Press, Princeton
2. Silver F H 1987 Biological materials: structure, mechanical properties, and modeling of soft tissues. New York University Press, New York
3. Currey J D 1979 Changes in the impact energy absorption of bone with age. Journal of Biomechanics 12: 459–469
4. Black J 1976 Dead or alive: the problem of *in vitro* tissue mechanics. Journal of Biomedical Materials Research 10: 377–389
5. Ker R F, Alexander R McN, Bennett M B 1988 Why are mammalian tendons so thick? Journal of Zoology 216: 309–324
6. Alexander R McN 1988 Elastic mechanisms in animal movement. Cambridge University Press, Cambridge
7. Cooper R R, Misol S 1970 Tendon and ligament insertion. A light and electron microscopic study. Journal of Bone and Joint Surgery 40A: 419–434
8. Mow V C, Kuei S C, Lai W M, Armstrong C G 1980 Biphasic creep and stress relaxation of articular cartilage in compression: theory and experiments. Journal of Biomechanical Engineering 102: 73–84.
9. Kempson G E, Muir H, Pollard C, Tuke M 1973 The tensile properties of the cartilage of human femoral condyles related to the content of collagen and glycosaminoglycans. Biochimica et Biophysica Acta 297: 456–472
10. Dorrington K L 1980 The theory of viscoelasticity in biomaterials. In: Vincent J F V, Currey J D (eds) The mechanical properties of biological materials. Symposia of the Society for Experimental Biology 34. Cambridge University Press, Cambridge, pp 371–382
11. Broom N D 1986 The collagenous architecture of articular cartilage: a synthesis of ultrastructure and mechanical function. Journal of Rheumatology 13: 142–152