Acta orthop. scand. *50*, 139–143, 1979

NON-LINEAR PROPERTIES OF DIAPHYSEAL BONE

An Experimental Study on Dogs

P. NETZ, K. ERIKSSON & L. STRÖMBERG

Department of Orthopaedic Surgery, Karolinska Hospital; Department of Strength of Materials and Solid Mechanics, The Royal Institute of Technology; Department of Experimental Surgery, Karolinska Institute; and Department of Surgery, Karolinska Hospital, Stockholm, Sweden.

The deformation of entire long bones from dogs has been studied in torsion. The load-deformation relationship is non-linear prior to ultimate failure and repeated successively increased loading and unloading in the non-linear range is found to reduce the stiffness of the bone. The non-linearity of the curve is assumed to be due to a change in the geometry of the bone originating from formation of small cracks in the cortex.

The bone material of entire long bones from dogs seems to be linearly elastic for all torques.

Key words: diaphyseal bone; elastic properties of bones; experimental surgery; stiffness of bone; torsional testing

Accepted 3.x.78

If you twist a diaphyseal bone to fracture you will find, when examining the broken parts, that they fit together and that no residual deformation of the parts can be detected.

These observations are in accordance with the opinion of surgeons experienced in open reduction of spiral fractures, i.e., that exact reduction of fractured ends is possible clinically and that no adjustment of their shape is required to reproduce the original bone geometry after healing (Müller et al. 1970). Thus the deformation and fracture behaviour of diaphyseal bone is typical of an elastic-brittle material.

Diaphyseal bones tested in torsion show however a non-linear torque-twist relationship prior to fracture. Many investigators have assumed the non-linearity to correspond basically to plastic deformation of the bone (Burstein et al. 1972, Pierkarski 1970, Chamay 1970, Pope & Outwater 1972).

The discrepancy between the clinical observations and the experimental results encouraged us to study the deformation of diaphyseal bone especially in the non-linear range. To gain insight into the nature of the non-linearity, repeated loading and unloading torsional experiments have been performed on diaphyseal bones.

MATERIAL AND METHODS

The test material chosen was seven related pairs of fresh canine diaphyseal bones, with closed

Financial support was obtained from the Swedish Medical Research Council (project no. B 77-17X-04788-02) and from the Trygg-Hansa Research Fund for Bodily Injury in Personal Accident Insurance, Stockholm, Sweden.

0001–6470/79/020139–05$02.50/0

epiphyseal lines, from long legged dogs of varying breeds and of both sexes. One bone of each related pair was selected as a control bone and the other was used as a test bone.

Preparation

The animals were sacrificed with lethal doses of mebumal sodium (Nembutal®, Abbott lab., USA) administered i.v. Immediately after sacrifice all soft tissues were removed, but the periosteum was kept intact, and the bones were immersed in a +37°C saline solution. Both bones of a related pair were tested within 30 minutes after sacrifice of the corresponding animal. At most, 10 minutes elapsed between the tests of the two bones of a related pair. All related pairs of bones were treated in the same manner.

Equipment

The test equipment and method has been described by Strömberg & Dalén (1976a). To permit reversal of the direction of twist at a predetermined torque, the equipment was provided with an additional device. In short, across the display of the time base recorder, on which the torque-twist curve is plotted, a light beam runs parallel to the twist axis at the predetermined torque. The light beam is connected to the twist machine over a switch. When the recorder pen breaks the light beam, reversion of the rotation direction is triggered by the switch. During a test the torque was determined as a function of the applied twist for the diaphyseal bone.

It has been shown that the maximum torque capacity of diaphyseal bone is independent of the angular velocity in the range 3–12 degrees per second (Strömberg & Dalén 1976a). In view of this the angular velocity in the present work was chosen to be 6 degrees per second. The angular velocity of the twist machine is constant and independent of torque within the test range.

Test procedure

The recorded torque-twist curve of torsionally tested diaphyseal bones comprises two characteristic parts (Figure 1); one linear starting from zero load, which at a critical torque (here called M_c) changes to a non-linear part prior to the ultimate fracture of the bone (at the torque M_f).

To study the properties of the bone material and especially the nature of the non-linear behaviour the procedure was as follows: The torque-twist master curve for a related pair was obtained by twisting one of the bones to ultimate fracture. On this curve the points corresponding to M_c and M_f were marked. The other bone, the test bone, was repeatedly loaded and unloaded in torsion. In the first load cycle the bone was loaded to a torque M_i slightly greater than M_c and subsequently unloaded. Loading to a torque M_i and unloading from M_i to zero torque, where M_i was increased slightly in every load cycle, was repeated until the bone finally fractured. Loading and unloading torque twist curves were registered for all load cycles.

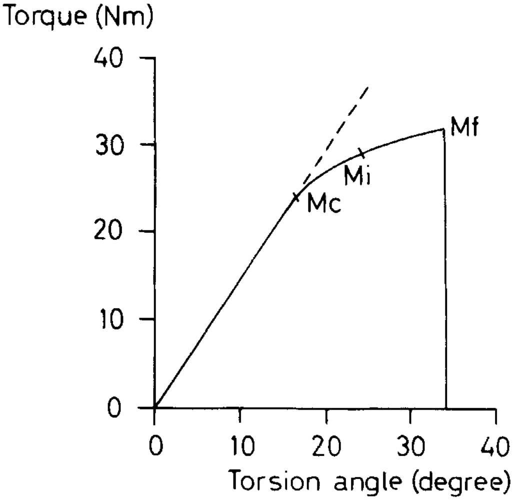

Figure 1. Example of a torque-twist curve for an entire long bone of a dog. Load is expressed in newtonmeters (Nm) and the twist in degrees.

M_c represents the critical point where the linear phase turns into the non-linear phase. M_i is a randomly chosen point on the curve in the non-linear range. M_f is the point on the curve where the ultimate failure occurs.

Calculation

The interesting feature of the load cycles is the slope of the linear part of the loading phases, or the stiffness of the bone. The stiffness is expressed as torque per unit twist (newtonmeters (Nm) per degree).

Conventional statistical methods were used to determine the significance of differences and correlations between the recorded parameters. $P < 0.05$ was chosen to indicate significant correlation and difference.

Table 1. The stiffness of paired long bones from seven dogs. Stiffness is expressed in torque per unit twist (newtonmeters per degree)

Dog no.	Stiffness of paired bones Nm/degree Left	Right	Percentage deviation
1	2.19	2.21	−0.9
2	1.00	1.01	−1.0
3	1.94	1.92	+1.0
4	1.20	1.18	+1.7
5	1.55	1.54	+0.6
6	2.01	2.06	−2.5
7	1.39	1.36	+2.2

The two bones of a pair were twisted in the same direction with respect to the geometry. One of the bones was twisted to ultimate failure and the other of the pair to a point M_i in the non-linear range of the recorded curve (see Figure 1). The slope of the linear phases was calculated and the percentage deviation from the mean value of the related pair is given. All tested bones had the same stiffness as their respective controls (mean = +0.2 per cent; s.d. = 1.7 per cent; $P > 0.05$).

Table 2. The stiffness after varying numbers of load cycles for one bone of a pair of long bones from seven dogs. Stiffness is expressed in torque per unit twist (newtonmeters per degree)

Dog no.	Stiffness (Nm/degree) Load cycle no. I	II	III	IV	Percentage deviation
1	2.19	1.96	1.65		−24.7
2	1.00	0.92	0.83		−17.0
3	1.94	1.82			− 6.2
4	1.18	1.13	1.04		−11.9
5	1.54	1.40			− 9.1
6	2.06	1.86	1.71	1.67	−18.9
7	1.36	1.24	1.20		−11.8

The deviation of the stiffness from the first loading and the last (where the ultimate fracture occurs) is given in per cent of the value of the first loading. In all cases the stiffness was smaller in the last loading compared with that of the first one (mean = −14.2 per cent; s.d. = 6.3 per cent; $P < 0.01$).

RESULTS

1) The slope of the linear part of the first loading curve was equal to that of the master curve for any related pair of bones (Table 1) ($P > 0.05$).
2) The slope of the linear part of the loading curves decreases as the load cycle number increases (Table 2) ($P < 0.01$).

DISCUSSION

It is known that drying or freezing affects the mechanical properties of biological materials (Smith & Walmsley 1957, Hirsch & Sedlin 1966, Strömberg & Dalén 1976b). This fact limits the scope of the present study to fresh material only. With the method used it is possible to perform testing within a few minutes after detaching the material from the donor (Strömberg & Dalén 1976a).

A previous study has shown that the torque-twist curves for both bones of a related pair of diaphyseal bones are almost equal and independent of the combination of directions of twist for the two bones (Netz et al. 1978). This means that for example, stiffness and maximum torque capacity of both bones are the same and this permits the use of one bone of a related pair as a control bone to determine the torque-twist curve for the pair.

Stress-strain relationships for materials are in practice obtained by inference from a load-deformation relationship for a body of the material considered. The two relationships are however identical (in normalized units) only if continuity of the body is preserved during the deformation cycle. Cracking, cavity formation and other phenomena by which new surfaces are formed, affect the load-deformation relationship of a body but not the properties of the material (Broek

1974). The stress-strain relationship for a material susceptible to cracking can therefore not be uncritically inferred from a load-deformation relationship.

Cracking, or localized fractures in combination with deformation implies a restriction upon the choice of specimen. Specimens containing only some part of the entire bone will suffer both from a different geometry and different test boundary conditions, quite apart from effects of specimen preparation upon mechanical properties (Hirsch & Sedlin 1966). As the load-deformation relationship is dependent upon the geometry of the body, in particular the non-linear part of the load-deformation curve is characteristic of the behaviour of the particular specimen only.

As regards deformation in combination with localized fracture the effect of size and shape of the specimen upon its non-linear properties cannot be excluded. It is our opinion that the deformation and fracture process of diaphyseal bone can be adequately studied only by using entire bones as specimens.

Bone material is known to be viscoelastic, i.e. its mechanical properties are time-dependent (Burstein & Frankel 1968, Currey 1970, McElhaney & Byars 1965). A preparatory test comprising loading and unloading of a test bone within the linear range of the corresponding master curve revealed, however, that the width of the hysteresis loop along the twist axis is insignificant compared to the elastic twist range. In fact, for the actual loading rate viscous effects can hardly be detected on a twist axis scale suitable for plotting the non-linear behaviour of the entire bone. This means that viscous effects within the linear range are insignificant compared to the effects discussed in the following.

The slope of the loading curves decreases as the load cycle numbers increase with a concomitant gradually increased load (Figure 2). The same result is also obtained if a complete loading and unloading cycle is considered a hysteresis loop: the mean slope of the loop decreases as the load cycle number increases. This means that the stiffness of the bone must have been affected. The reduction of the stiffness is to such an extent that the non-linear behaviour of the bone can be explained essentially by this effect.

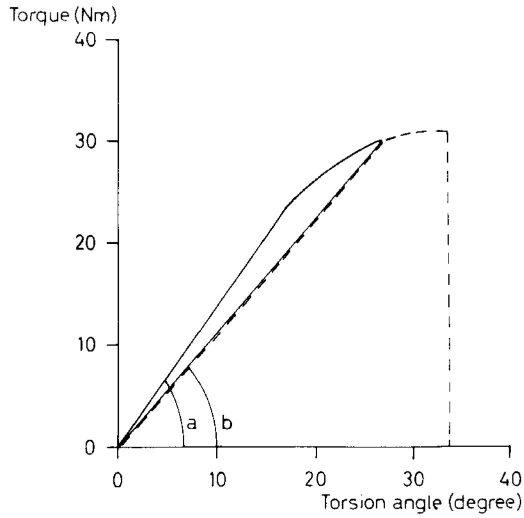

Figure 2. A schematic torque-twist curve on repeated and successively increased loading, showing the decrease of the slope of the linear phase - stiffness. Stiffness is expressed in torque per unit twist (newtonmeters per degree).

The first loading is represented by the continuous black line. The second loading, which goes to ultimate failure is represented by the broken line and the linear phase of this loading forms the angle b. In the present experiments the angle a was greater than the angle b.

The change in the stiffness requires a corresponding change in the bone geometry. A change in the geometry of the bone can be explained by assuming that small cracks have been formed in the bone material as a result of the deformation in the non-linear range.

Considering the structure of the bone material and the stress distribution in an uncracked bone it is reasonable to assume that the crack plane is perpendicular to the bone surface and parallel to its longitudinal axis (similar to those described by Frost 1960). From theoretical considerations it is easily concluded that cracks with the proposed orientation reduce the twist stiffness of the bone. A quantitative calculation is however a lengthy task. In order to approximately illustrate the effect of axial cracks upon the twist stiffness of a cylinder qualitatively, thin-walled cylinders of an elastic material with axial slots through the wall were twisted in torsion. Effectively, the twist stiffness decreased as the number and length of the slots increased. We consider this effect as an additional indication of the formation of microcracks in tested diaphyseal bones, during the non-linear range in torsion.

We are therefore tempted to suggest that bone material is linearly elastic for all torques, that is, even in the non-linear range and that the non-linearity of the torque-twist curve prior to fracture of the bone is due to the formation and growth of a number of small cracks in the bone material. Microscopic inelastic effects, e.g. as observed by Chamey (1970) in the immediate neighbourhood of a fracture surface, are not excluded by this suggestion.

We assume that the torsional fracture process of diaphyseal bone is a two-stage process. In the first stage, occurring during the non-linear phase, an increasing number of small cracks in the cortical layer are formed. The bone still bears the applied torque but its capacity to withstand an increasing torque is reduced. In the second stage, at the maximum torque, ultimate failure of the bone is brought about by sudden propagation of small cracks forming the spiral type of fracture.

REFERENCES

Burstein, A. H. & Frankel, V. H. (1968) The viscoelastic properties of some biological materials. *N.Y. Acad. Sci. (Wash.)* **146,** 158–165.

Burstein, A. H., Currey, J. D., Frankel, V. H. & Reilly, D. T. (1972) The ultimate properties of bone tissue: the effect of yielding. *J. Biomech.* **5,** 35–44.

Broek, D. (1974) *Elementary engineering fracture mechanics.* Noordhoff Int. Publ., Leyden, Netherlands.

Chamey, A. (1970) Mechanical and morphological aspects of experimental overload and fatigue in bone. *J. Biomech.* **3,** 263–270.

Currey, J. D. (1970) The mechanical properties of bone. *Clin. Orthop.* **73,** 210–231.

Frost, H. M. (1960) Presence of microscopic cracks in vivo in bone. *Henry Ford Hosp. Bull.* **8,** 25–35.

Frost, H. M., Roth, H. & Villanueva, A. R. (1961) Physical characteristics of bone. *Henry Ford Hosp. Bull.* **9,** 163–170.

Hirsch, C. & Sedlin, E. D. (1966) Factors affecting the determination of the physical properties of femoral cortical bone. *Acta orthop. scand.* **37,** 29–48.

McElhaney, J. H. & Byars, F. (1965) Dynamic response of biological materials. ASME paper 65-WA/HUF 9.

Müller, M. E., Allgöver, M. & Willenegger H. (1970) *Manual of internal fixation.* Springer-Verlag, Berlin-Heidelberg-New York.

Netz, P., Eriksson, K. & Strömberg, L. (1978) Torsional strength and geometry of diaphyseal bone. *Acta orthop. scand.* **49,** 430–434.

Piekarski, K. (1970) Fracture of bone. *J. appl. Physiol.* **41,** 215–223.

Pope, M. H. & Outwater, J. O. (1972) The fracture characteristics of bone substance. *J. Biomech.* **5,** 457–465.

Sammarco, J. G., Burstein, A. H., Davies, W. L. & Frankel, V. H. (1971) The biomechanics of torsional fractures. The effect of loading on ultimate properties. *J. Biomech.* **4,** 113–117.

Smith, J. W. & Walmsley, R. (1957) Factors affecting the elasticity of bones. *J. Anat. (Lond.)* **93,** 503–523.

Strömberg, L. & Dalén, N. (1976a) Experimental measurement of maximum torque capacity of long bones. *Acta orthop. scand.* **47,** 257–263.

Strömberg, L. & Dalén, N. (1976b) The influence of freezing on the maximum torque capacity of long bones. An experimental study. *Acta orthop. scand.* **47,** 254–256.

Correspondence to: Pelle Netz, Department of Orthopaedic Surgery, Karolinska Institute, S-10401 Stockholm 60, Sweden.