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The diaphyseal bone under torque

An experimental study on dogs

BY

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References to the above-mentioned papers are made in the text with the Roman numerals I - IV.

CONTENTS

INTRODUCTION	5
AIMS OF THE PRESENT STUDY	7
MATERIAL	9
METHODS AND TEST EQUIPMENT	10
TEST PROCEDURE	12
CALCULATIONS	14
STATISTICAL METHODS	16
RESULTS	17
DISCUSSION	21
CONCLUSIONS	27
ACKNOWLEDGEMENTS	28
REFERENCES	29

INTRODUCTION

Living biological materials possess to a certain extent the unique ability to adapt themselves to functional demands. This adaptability of the diaphyseal bone was recognized by Wolff in 1870. In order to understand the remodeling of the diaphyseal bone, its mechanics have been the subject of increasing scientific interest. The literature in the field is extensive and the terminology sometimes confusing and inconsistent. For this reason certain mechanical terms used in the present study in accordance with general usage in engineering are briefly defined below.

Force, which is an axiomatic concept in classical mechanics, is best described in technical terms, although the simple expressions "push", "pull", and somewhat carelessly, "twist" may serve as illustrations of our everyday idea of the concept. Mathematically, forces are vectors, i.e. they have both magnitude and direction. Common technical terms are tensile forces, which increase the distance between two points in a body on a line in the direction of the force, compressive forces opposite to tensile, and shearing forces which displace one layer with respect to another parallel to the plane of the layers. Strain is the relative displacement of one point in regard to another. Strains can be tensile, compressive or shear. Stress is force per unit area on an imagined surface in a body. Stresses can also be tensile, compressive or shear. In a cylinder, twisted around its longitudinal axis, the stress and strain state is shear on planes a) perpendicular to the longitudinal axis and b) in the longitudinal axis, and tensile respectively compressive on planes 45° to the longitudinal axis. The force which twists a body is called torque. The load necessary to displace the load point a unit length in the direction of the load is called the stiffness of a body and is expressed in force per unit length. For a twisted body the stiffness is the torque per unit twist (Fig. 1).

The most important types of deformation of biological materials are elastic, plastic and viscous deformation. Time-independent deformation which disappears upon release of load is called elastic deformation. A body is linear elastic if its deformation is proportional to the applied load. Correspondingly, a material is linear elastic if the stress-strain relationship is linear. The ratio stress per unit strain (tensile or compressive), is called Young's modulus of the material. Most materials are elastic for small loads.

For materials subjected to an increasing stress, there may exist a point (before fracture) in the stress-strain relationship from which, on release of stress, the strain no longer returns to its original state. The stress at which this phenomenon begins is called the elastic limit of the material. If the remaining deformation is independent of time the material is plastic, or time-independent deformation which remains upon release of load is called plastic deformation. A material which after a certain amount of elastic deformation deforms plastically is called elastic-plastic and the elastic limit the yield point of the material. Time-dependent deformation of a loaded or unloaded body is called viscous deformation.

The fracture process is most simple in elastic materials. The load-deformation relationship is linear to fracture and once fracture is initiated propagation follows even if the load is released. This is due to the fact that the elastic energy stored in the body during loading is reversible, i.e. immediately available for propagation of fracture. The fracture of glass and porcelain illustrates the fracture of an elastic material.

The fracture of an elastic-plastic material is preceded by dissipation of energy through plastic deformation. In metallic materials plastic deformation is associated with generation and motion of dislocations, a feature charac-

teristic of the lattice structure of metals. Compared to the time for fracture to propagate a unit distance in an elastic material, plastic deformation in a region of equal size in an elastic-plastic material is a time-consuming process. In metals which are mainly plastic, e.g. lead, the fracture propagation speed is slow and will stop immediately upon release of load.

In many previous experimental studies the bone material was postulated to exhibit plastic deformation in macroscopical parts of the bone (Currey 1970, Piekarski 1970, Burstein et al. 1972a, 1973, Engesaeter et al. 1978). However, considering the diaphyseal bone twisted to fracture, it is well known to the experienced surgeon that exact reduction of the fracture ends is possible and fixation can be accomplished without any residual macroscopical deformation of the entire bone (Olerud & Johansson 1967) i.e. bone material seems elastic to the surgeon.

The discrepancy between the postulated behaviour and clinical experience initiated the present study.

The aims of this work were:

- 1) To study each phase of deformation of the entire canine diaphyseal bone under torque.
- 2) To study the influence of anatomical asymmetry on the maximum torque capacity and on the torsional stiffness of entire canine diaphyseal bones.
- 3) To study the nature of the non-linear deformation of the entire canine diaphyseal bone under torque.

Torque (Nm)

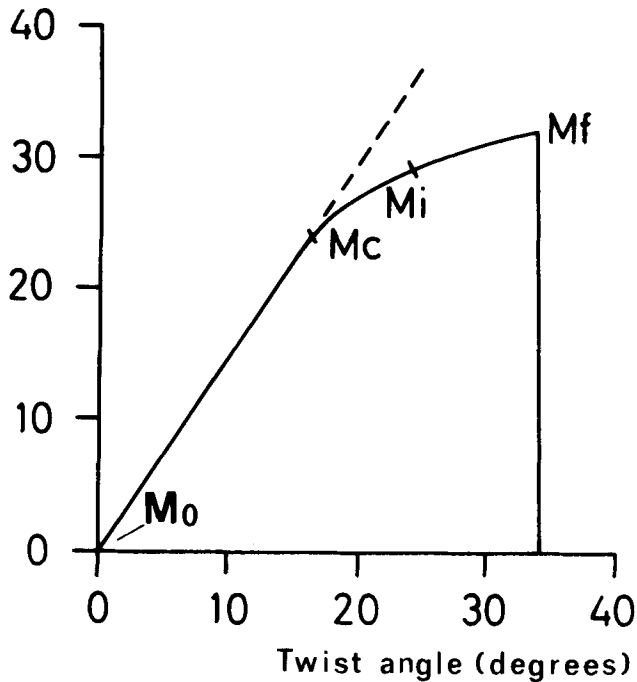


Figure 1. Example of a load-deformation relationship in torsion for an entire canine diaphyseal bone. Load is expressed in newtonmeters (Nm) and twist in degrees.

M_0 is the point at zero torque and twist, M_c the critical point where the linear phase turns into the non-linear phase, M_i a randomly chosen point in the non-linear range of the curve and M_f is the point where the ultimate failure occurs. The slope of the linear part of the curve ($M_0 - M_c$) represents the stiffness of the material.

MATERIAL

In all tests, diaphyseal bones from healthy adult long legged dogs of varying breeds and of both sexes were used.

To test the influence of the anatomical asymmetry on the strength and stiffness of diaphyseal bones twenty related pairs of previously frozen and thawed femora and tibiae were chosen (I).

To study the non-linear properties of canine diaphyseal bones under torque, seven related pairs of fresh femora and tibiae were tested (II).

The test material, used to study formation and/or growth of cracks and its relationship to non-linear behaviour of diaphyseal bone under torque, consisted of nine fresh femora and tibiae (III).

In the study of the ultimate failure, two healthy animals were selected from a total of thirty dogs. Two fresh related pairs of femora and tibiae were tested (IV).

Sacrifice

The dogs were sacrificed with i.v. lethal doses of mebumal-sodium (Nembutal^R, Abbott Laboratories, North Chicago, Ill, USA).

Bone preparation

Preparation of the bones started immediately after sacrifice of the animal. All soft tissues were removed in study III, but in the remaining studies the periosteum was left intact. During the preparation, the bones were kept humid with physiological saline solution. Both bones of a related pair were always treated in the same way (I, II, IV).

The bones intended for studies of the influence of anatomical asymmetry on maximum torque capacity and stiffness (I)

were wrapped in gauze saturated with physiological saline solution and placed into plastic bags. After sealing the bags were stored at a temperature of -30°C . Before testing the bags were immersed into physiological saline solution at $+20^{\circ}\text{C}$ for two hours.

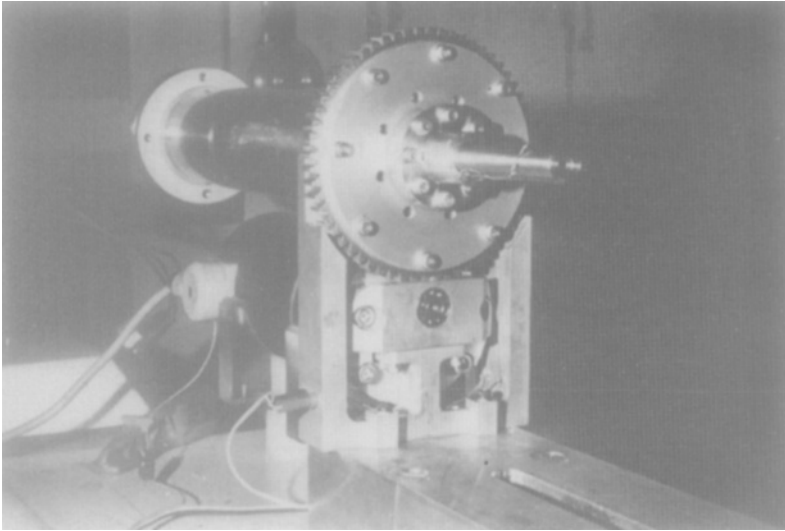


Figure 2. The torsion machine with the entire canine diaphyseal bone mounted in the plexiglass chamber.

METHODS AND TEST EQUIPMENT

Torque and stiffness measurements (I-IV)

In the present investigation the maximum torque capacity and the stiffness of entire canine diaphyseal bones were measured by means of a method and torsion machine described by Strömberg & Dalén (1976a).

The entire diaphyseal bones were fixed in the machine grips by casting the bone ends in two metal cuffs. The castings consisted of a metal alloy with a solidification temperature of 47.5⁰C. By this technique undue stress concentrations in the bone ends were prevented and rendered possible testing of the bone within 15 - 30 minutes after sacrificing the animal.

The torsion machine permits twisting of the test bone at a constant twist rate, irrespective of the load within the test range (Fig.2). It has been shown that the maximum torque capacity is independent of the angular velocity in the range 3 - 12 degrees per second (Strömberg & Dalén, 1976a). Hence, the angular velocity in the present study was fixed at 6 degrees per second. The error of the method, when measuring maximum torque capacity, was 3% (Strömberg & Dalén, 1976a).

The torque as a function of time was registered on a time base recorder (Hewlett & Packard 7005 B; I & II) or on a multichannel light beam recorder (Ultralette 5651, ABEM, Bromma, Sweden; III & IV).

Loading and unloading measurements (II)

To permit reversal of the twist direction at a predetermined torque, the equipment was provided with an additional device. Across the display of the time base recorder, on which the torque-twist curve was plotted, a light beam ran parallel to the twist axis at the predetermined torque. The light beam was connected to the twist machine over a switch. When the recorder pen broke the light beam, reversion of the rotation direction was triggered by the switch.

Acoustic emission (III)

The most common types of mechanical behaviour of material under applied loads are deformation and fracture. However, many other phenomena, such as dislocation generation and

motion, crack formation and growth, may also occur. A typical feature of these phenomena is dissipation and redistribution of the energy stored in the material. The redistribution of energy is associated with propagation of stress waves in the material, which are detectable with the acoustic technique (Dunegan & Harris 1969). Acoustic detection of phenomena in materials is called acoustic emission or A.E. technique.

The stress waves in a material are detected by a transducer applied on the surface of the body. The transducer converts strains associated with the stress waves to electric signals which are plotted after processing.

In the acoustic emission study a sensing transducer (Type D 140 A, Dunegan/Endevco, San Juan Capistrano, Calif., USA) connected to a preamplifier, filter and processor unit with a digital recorder, a reset clock and an audiomonitor, was used (type 801 P, 202, 301, 402, 702, respectively, Dunegan/Endevco). The A.E. signals within the range of 140 - 500 kHz were registered and the transducer had a maximum sensitivity at 220 kHz.

High speed filming (IV)

The final fracture process of canine diaphyseal bone under torque was filmed with a high speed camera equipment (HYCAM K20S4AW, I. Weinberger, Zürich, Switzerland). The film speed was 5 000 exposures per second, i.e. 120 meters of 16 mm film were consumed in 3.5 seconds. Due to a necessary acceleration time only 2.5 seconds were available for registration.

TEST PROCEDURE

Throughout the test procedure the bones were kept humid with physiological saline solution at a temperature of + 37⁰C (II) or at approximately +20⁰C (I, III, IV). Prior

to each experiment the equipment was calibrated against a known torque.

From each related pair of diaphyseal bones, one was selected at random as test bone and the other bone as its control (I, II & IV).

Influence of anatomical asymmetry on strength and stiffness (I).

One bone, chosen at random, of a pair was twisted outwards and the other bone inwards. The maximum torque capacity and the stiffness were calculated.

Non-linear properties (II)

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 (Strömberg & Dalén 1976a, I). On this curve (Fig.1) were marked the points where the linear part of the curve turns non-linear (M_c) and corresponding to fracture (M_f). 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 of the control bone, and subsequently unloaded. Loading to a torque slightly greater than that in the preceding load cycle and unloading to zero torque (M_o) was repeated until the bone finally fractured.

Loading and unloading torque-twist curves were registered for all load cycles. The stiffness at each new loading and the maximum torque capacity at final fracture were calculated.

Formation and/or growth of cracks (III)

Prior to testing, the A.E. equipment was calibrated and the amplification adjusted to minimize extraneous noise, e.g. from the test machine. On the humid bone surface without periosteum the transducer was applied and great care was taken to minimize transmission losses of the acoustic signals.

During testing the bones were twisted inwards to fracture and both torque and A.E. counts were simultaneously plotted as functions of applied twist on the multichannel light beam recorder. Furthermore, the A.E. signals were checked audibly with the audiomonitor.

Ultimate failure (IV)

The control bone of a pair was first twisted to ultimate failure and the maximum torque capacity determined. Subsequently the test bone was twisted to ultimate failure. The onset of the filming period was guided by the previously obtained maximum torque capacity of the control bone (Strömberg & Dalén, 1976a, I). To avoid the effects of lighting heat, the bones were continuously sprayed with cold physiological saline solution during the filming.

CALCULATIONS

Maximum torque capacity (I, III & IV)

The maximum amplitude of the recorded curves, the point Mf (Fig.1), was measured. By means of the calibration data the maximum torque capacity, expressed in newtonmeters (Nm), was determined. The difference between the maximum torque capacity of the right and left bone of a pair was expressed in per cent of the mean value of the two bones (I).

Stiffness (I & II)

The slope of the linear part of the recorded torque-twist relationship, represents the stiffness of the intact diaphyseal bone (Fig.1). The bone stiffness was expressed as torque per unit twist (newtonmeters per degree) of the linear part.

The difference between the stiffness of the right and left bone of a pair was expressed in per cent of the mean value of the two bones. In study II the difference of the stiffness between the first and the last loading was expressed in per cent of the value of the first loading.

Torsional deformation (I, II & III)

The torsional deformation is calculated to start where the tangent to the elastic part of the torque-twist curve crosses the abscissa. The point on the torque-twist curves where the curve turned non-linear (M_c , fig.1) was determined (II & III).

The twist range was in study III divided in intervals of 1.2 degrees which corresponds to 0.2 seconds. The interval containing M_c was determined and was called "turnover" interval.

Formation and/or growth of cracks (III)

Characteristical A.E. signals for crack formation and/or growth appear on the recorded curves step-wise and with increased frequency towards the ultimate failure of the bone. Furthermore, the signals when audibly checked come in spurts. These characteristics differ from the smoother signals decreasing towards ultimate failure which are typical for dislocation generation and motion.

The deformation intervals during which any acoustic activity characteristic for crack formation and/or growth occurred were noted from the recorded curves. Acoustic activity not characteristic for crack formation and/or growth was disregarded.

Fracture propagation speed (IV)

The high speed film exposures of the final fractures were analyzed. A comparison between the length of the fracture path on consecutive exposures was made. Half the length of the final fracture was measured on the fractured bones. By using the film speed, a lower limit of the fracture propagation velocity was calculated.

STATISTICAL METHODS

Conventional statistical methods were used to determine mean values and standard deviations. Statistical significance of differences between the recorded parameters was calculated by using the paired t-test (I & II) or the sign test (III) $P < 0.05$ was chosen to indicate significant difference.

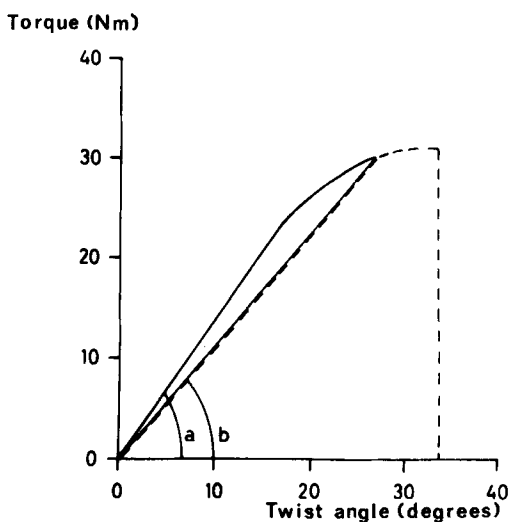


Fig. 3. A schematic torque-twist curve in loading and unloading experiment with a successively increased load. The slope of the linear phase, representing the stiffness of the bone, is decreased at every new loading. The first loading is represented by the continuous black line. The second loading, which results in final fracture of the bone, is represented by the broken line and the linear phase of this loading forms the angle b. Angle a in this study was always greater than the angle b.

RESULTS

At all torsional tests a torque-twist curve was recorded comprising two characteristic parts (Fig.1). Starting from zero load (M_0), the recording was linear and at a critical point (M_c) turned non-linear prior to the final fracture (M_f).

Influence of anatomical asymmetry on strength and stiffness (I)

No significant effect was found of the direction of twist upon the maximum torque capacity or upon the stiffness of related pairs of diaphyseal bones twisted in opposite directions with respect to the bone geometry. (Maximum torque capacity: Mean= 0.4 per cent, S.D. 6.2 per cent. Stiffness: Mean= 1.3 per cent, S.D. 5.1 per cent.)

Non-linear properties (II)

The slope of the linear part of the first loading curve did not significantly differ from that of the master curve for the corresponding control bone (mean= 0.2 per cent, S.D. 1.7 per cent).

The slope of the linear part of the loading curves decreases as the load cycle number increases (Fig.3) A significant deviation was found upon the stiffness of the first and the last loading expressed in per cent of the value of the first loading (mean= -14.2 per cent, S.D. 6.3 per cent $P < 0.01$).

Formation and/or growth of cracks (III)

The linear part of the torque-twist curve was within the range ~ 30 - 50 per cent of the maximum twist.

The number of A.E. signals characteristic for crack formation and/or growth, was significantly higher in the non-linear range compared with the linear range of the torque-twist curve ($P < 0.01$). In fact, characteristic signals for crack formation and/or growth were registered solely in

the non-linear range. An example of a registration is shown in Fig. 4.

For bones with the transducer close to the fracture propagation path, the onset of the acoustic signals was concomitant with the beginning of the non-linear range on the recorded torque-twist curves. The onset of acoustic signals was later when the transducer was placed away from the fracture propagation path.

The intensity of the recorded stress waves proved to be dependent on the smallest distance between the transducer and the fracture propagation path. It was impossible to determine the quantitative relationship between the two parameters, but the shorter the distance the more intense the recorded stress waves.

Fracture propagation speed (IV)

Exposures immediately before and after the final fractures show that the fractures were accomplished in the interval between two consecutive exposures i.e. in 0.0002 seconds. The calculated lower limit of the fracture velocity was 340 meters per second (Table I). The intense lighting on the test bones did not seem to affect the maximum torque capacity (Table I) or the type of the final fracture. The bones failed through spiral fractures.

BONE	MAXIMUM TORQUE CAPACITY (Nm)		FRACTURE PROPAGATION SPEED (m/s)
	Control bone	Test bone	Test bone
Tibia	26.2	26.8	› 385
Femur	28.2	27.8	› 340

TABLE I. The maximum torque capacity and the calculated lower limits of the fracture propagation speed in canine diaphyseal bones. The maximum torque capacity is given in newtonmeters (Nm) and the fracture propagation speed in meters per second (m/s).

DISCUSSION

The diaphyseal bone is a heterogeneous construction of materials. The distribution of spongy and cortical bone varies in different parts of the bone as well as the inhomogeneities, such as blood vessel channels, intracortical cavities etc. (Evans & Lebow 1951, Currey 1962, Piekarski et al. 1970, Robertson et al. 1978, Robertson & Smith 1978) This fact limits the use of machined bone specimens when studying the mechanical properties of entire diaphyseal bones. Specimen preparation, being generally time-consuming, promotes postmortal changes, which affect the mechanical properties of the bone. It has previously been shown (Evans & Lebow 1951, Smith & Walmsley 1957, Strömberg & Dalén 1976b) that drying, heating, freezing and thawing significantly affect the strength and stiffness of the cortical bone. Furthermore, specimen preparation may produce micro-damage to the material. The surface preparation of the specimens affects to a great extent the mechanical properties of the test body and discontinuities on the surface will totally change the strength of the specimen (Swanson et al. 1971).

The load-deformation relationship and especially its non-linear part is affected by the geometry of the test body and is thus particular for the specimen type (Broek 1974). The mechanical properties (e.g. the maximum torque capacity) of symmetrical bones have been studied by many authors and the use of one bone of a pair as a test bone and the other as its control is justified (Mather 1967, Burstein & Frankel 1971, Strömberg & Dalén 1976a), provided that the test bodies are treated and tested in the same way. In view of this, it is my opinion that the deformation and the fracture process of diaphyseal bone is advantageously studied only by using entire bones as test bodies.

The method used in the present study has been well analyzed and tested on entire canine diaphyseal bones since

1976 (Strömberg & Dalén). Testing is possible within a few minutes after detaching a bone from the donor, thus avoiding any significant postmortal changes. This fact favours the use of fresh material only, but after careful consideration previously frozen bones were preferred in one study (I). The two bones of a pair were always treated in the same way and in study I only comparative studies between the two bones of a pair were made.

Bone material is known to be viscoelastic (MacElhaney & Byars 1965, Burstein & Frankel 1968, Currey 1970, Sammarco et al. 1971, Wright & Hayes 1977). A preparatory loading and unloading test within the linear range of the master curve revealed, however, that the width of the hysteresis loop along the twist axis is negligible compared with the elastic twist range of the bone (c.f. Hirsch 1969). This means that viscous effects are insignificant compared with those discussed in the following.

To judge from the available literature, great interest has been paid to the anatomic asymmetry of paired diaphyseal bones, considered as mirror images of each other, i.e. the twist direction with respect to the bone geometry has been the same for both bones (Sammarco et al. 1971, Strömberg & Dalén 1976 a+b). The possible effect of the twist direction upon the torsional strength has thereby been eliminated. Nothing contradicts the assumption that the torsional strength of an asymmetric body depends upon the direction of twist.

In the study (I) which deals with maximum torque capacity and stiffness of bones of related pairs twisted in different directions with respect to the bone geometry, no differences could be found in the investigated parameters. The geometrical asymmetry is obviously compensated for by a corresponding asymmetry of the internal bone architecture, or the differences are too small to be detected with the method used in this study. The compensation of the anat-

mical asymmetry is in accordance with the principle of adaptation to functional demands (Wolff's law) and further indicates that the bone material in the diaphyseal bone is heterogeneous (Wolff 1870).

The stiffness of the cortical bone was by Smith & Walmsley (1957) shown to be affected by the temperature of the bone. They found that the value of Young's modulus was inversely proportional to the temperature, decreasing from 1.45 at +4⁰C to 1.25 at +37⁰C. In order to minimize changes of the elastic properties to be investigated, the temperature of the bones in study II was held at +37⁰C for all bones tested.

Studies of the stiffness of machined bovine specimens were carried out by Carter & Hayes (1977a,b). They found a marked decrease of the stiffness both in flexural and in tension fatigue. In study II the non-linear behaviour of entire diaphyseal bones was studied by means of cyclic torsional loading.

The slope of the loading curves decreased as the load cycle number increased. If a complete loading and unloading cycle is considered a hysteresis loop, the same result was obtained, i.e. the mean slope of the loop decreased as the load cycle number increased. This means that the stiffness of the bone has been affected. The change of the stiffness requires a change of the bone geometry (Broek 1974), which 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. Paper II thus demonstrates a probable geometrical change of the canine diaphyseal bone under torque and this finding is further analyzed and explained in the following (III).

Cracks have been observed in cortical as well as in spongy bone (Rutishauser 1951, Frost 1960, Frost et al. 1961,

Pugh et al 1973, Carter & Hayes 1977a). Cracks perpendicular to the bone surface and parallel to the long axis of the bone reduce the stiffness of the bone (Broek 1974). Piekarski (1970) and Carter and Hayes (1977b) suggested that what they call "the postyield regime" of compact bone is the result of structural damage. Carter & Hayes (1977a) provided very clear evidence of microdamage of compact bone specimens and suggested a relationship with non-linear deformation of the bone material.

By means of the acoustic emission (A.E.) technique, which is non-destructive, stress waves generated in entire diaphyseal bones twisted to final fracture were analyzed (III). The pattern of the A.E. stress waves depends to a large extent upon the generating process, e.g. the character of stress waves generated by dislocation differs from that of stress waves from cracking (Dunegan & Harris 1969). Briefly, processed signals associated with dislocations are smooth, whereas those associated with cracking appear stepwise and sharply (Tetelman 1971). The A.E. technique thus permits identification of the generating process.

A.E. signals characteristic for formation and/or growth of cracks appeared only in the non-linear range of the torque-twist curve. The variation of the frequency of the A.E. counts was noteworthy for different bones. The fewest A.E. counts were obtained with the transducer remote from the final fracture path. Pilot tests, however, have shown that ultrasonic sound in compact bone is severely damped when propagating in other directions than along the long axis of the bone (Sandberg 1978). This may be an effect of the internal architecture of the bone, e.g. the longitudinal orientation of the osteons (cf. Pope & Outwater 1974, Robertson & Smith 1978). Plastic deformation in an axially twisted cylinder engages a uniformly distributed large volume of material and the observed frequency of A.E. counts should in this case be fairly independent of the location of the transducer (Mendelson 1968). Even when the A.E. counts

characteristic for cracking were few, signals typical for dislocation did not intervene.

The application of the A.E. technique made it possible to confirm a previous assumption that the geometry of the diaphyseal bone changes due to microcracking when twisted in the non-linear range. Elastic-plastic and elastic-cracking materials may have identical non-linear loading relationships, although the nature of the material behaviour is totally different in the two cases (Broek 1974).

The microcracks are more effective local stress raisers than the inhomogeneities (Burstein et al. 1972b, Piekarski 1970, Currey 1962). Thus it is reasonable to assume that they are decisive for initiation and propagation of the final fracture.

The ultimate failure of entire canine diaphyseal bone under torsion was studied in paper IV. The final fracture process of the entire diaphyseal bone proved to be accomplished within 0.0002 seconds. The propagation speed of the final fracture is considerable and too high to permit details of the fracture process to be discerned with the equipment used in this investigation. In order to estimate a lower limit of the fracture propagation speed, it was assumed that the fractures propagate from one point in two directions. It is, however, possible that the final fracture is formed through joining of local fractures initiated at different points. The true propagation speed of a local fracture may thus be lower than the estimated value. Plastic deformation of macroscopical parts of the bone is a time-consuming process in comparison with that of failure. Hence only a very thin layer in the immediate neighbourhood of the fracture surface can be deformed plastically during the fracture process.

Robertson et al. (1978) studied the fracture toughness of

bovine bone material and found that the size of the plastic zone around the crack tip is less than 0.02 mm. On this basis, they concluded that crack propagation involves localized disruption around individual or small groups of lamellae. This observation is quite in accordance with a high fracture propagation speed. Plastic effects are too small to be detected macroscopically. They are not, however, ruled out but may occur prior to fracture around inhomogeneities and close to the fracture surface during fracture, although on a microscopical scale (Chamay 1970).

Thus the torsional fracture process of diaphyseal bone is a two-stage process. In the first an increasing number of small cracks are formed in the cortical layer. In the second, ultimate failure is accomplished by sudden propagation of small cracks at the maximum torque.

CONCLUSIONS

On the basis of the present investigation it is concluded that:

1. The maximum torque capacity and the stiffness of canine diaphyseal bones are both independent of the direction of twist, i.e. symmetrical properties. This is in accordance with the principle of adaptation to functional demands of the canine diaphyseal bone.
2. The material of canine diaphyseal bone is linearly elastic for all torques even in the non-linear range of the torque-twist relationship.
3. The non-linear deformation of canine diaphyseal bone prior to final fracture is caused by microdamage.
4. The final torsional fracture of canine diaphyseal bone takes place within 0.0002 seconds. This is incompatible with plastic deformation of macroscopical parts of the bone.
5. The fracture process of canine diaphyseal bone comprises two steps: firstly an increasing number of small cracks are formed during the non-linear deformation and secondly ultimate failure occurs by sudden propagation of the small cracks.

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