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Solid mechanics and strength of bone in young dogs

Fresh tibiae and femora from 22 harriers ranging in age from 8 to 44 weeks have been studied with a computerized torsion-machine. The growth of the dogs studied ceased at an age approaching 30 weeks. The strength and stiffness of the entire bones also ceased to increase at that age, and the increase in stress at fracture and shear modulus of the bone material strongly declined in impetus. This means that no important maturation, as measured in strength of the bone material, takes place after cessation of growth. The fracture twist angle per unit bone length was greatest for the younger bones and decreased to an age of around 25 weeks; the relation, however, between the linear and the non-linear part of the torque-twist curves did not change.
Furthermore, the study indicates that juvenile bones, like adult bones, are essentially elastic-brittle, and there was no sign of increased plasticity in the young bones.

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Fractures in young individuals are often quite different from fractures in adults (Blount 1955, Borden 1975). At torsional tests of mature diaphyseal bone, the load-deformation relationship is initially linear. Beyond a certain point, the torque-twist curve turns non-linear. This non-linear behaviour has been described as a plastic deformation of the bone (Piekarski 1970, Burstein et al. 1972), which means that a complete unloading of the bone prior to fracture will leave a permanent deformation.

In 1975, Currey & Butler showed a relatively longer non-linear deformation prior to ultimate fracture for juvenile bone as compared with adult bone, and concluded that juvenile bone deforms more plastically than adult bone.

Netz and coworkers (1979, 1980a) have demonstrated that the non-linear deformation of adult diaphyseal bone is not mainly caused by plastic deformation; they showed that a gradual formation and/or growth of microcracks successively decreases the stiffness of the bone.

We have now studied the mechanical properties of juvenile diaphyseal bone and age-dependent changes in its solid mechanics and relevant strength parameters.

Material and methods

Material

The tibiae and femora of 22 healthy harriers of both sexes from five related litters were used. The age of the animals varied from 8 to 44 weeks. The dogs were bred in the same kennels and received adequate food and exercise. They were controlled daily, and were weighed on the day of the experiment.

Methods

Preparation. Immediately after the dogs had been sacrificed with Pentothal-Natrium®, the paired femora and tibiae were removed. The bones were kept immersed in physiologic saline at 19–22 degrees C. They were immediately freed from all soft tissues, including the periosteum. Until torsional testing, the bones were wrapped in saline-soaked gauze.

Torsional tests. The metaphyses of the bones were fixed in a torsional test machine, as previously described by Strömberg & Dalén (1976). This torsion machine was supplemented with computerized controlling and recording functions. The bones were twisted with a constant angular velocity of 6 degrees per second. All data characterizing the torque-twist curve were saved on disc.

I. One of the bones of each pair was twisted inwardly to ultimate fracture.

The parameters given by the computer were:

1. Torque at fracture (Nm).
2. Twist at fracture (degrees).
3. Stiffness (Nm/degree).
4. Linear twist range (degrees).

II. The corresponding contralateral bone was twisted until it was just starting to deform non-linearly, and thereafter immediately unloaded to zero load by reversal of the direction of twist. After a 60-s pause at zero-load, the bone was loaded in the same direction as previously but to a deformation exceeding the previous one by 2 degrees, and thereafter unloaded again. This procedure, with a successively increased deformation at each load cycle, was repeated until ultimate fracture occurred (Figure 1). The values given by the computer were:

1. Torque at fracture (Nm).
2. Twist at fracture (degrees).
3. Stiffness at each load cycle (Nm/degree).
4. Residual deformation after every complete load cycle (degrees).

Morphological recordings. Prior to the test, the lengths of all the bones and their two smallest perpendicular diaphyseal diameters were measured with a sliding caliper. The length of the twisted part of the bone, i.e. between the castings, was also measured. The bones were wiped dry before weighing, and X-rays were taken to estimate the fusion of the epiphysis. The mean cortical thickness was calculated from 10 measurements along the path of the fracture.

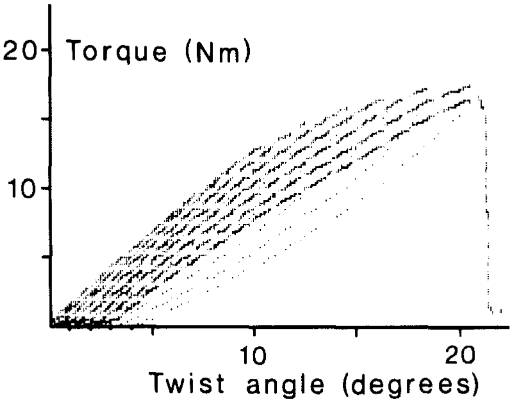

Figure 1. One of the bones from each tested pair has been repeatedly loaded and unloaded within the non-linear deformation. Twist has been successively increased at each new load-cycle until ultimate fracture. The residual deformation at the start of the fracture load-cycle was 3.0 degrees.

Calculations

Strength parameters. In preparatory torsional tests on entire tibiae and femora, it was found that cross-sections at the fracture site were almost circular and that the outer diameter of the bones (2b, see below) showed no significant variations in this context. This justified the following calculations.

The polar moment of inertia (I_p) is expressed as follows, in accordance with the formula for a hollow tube:

$$I_p = \frac{\pi}{2}(b^4 - a^4)$$

where 'b' (the outer radius) equals 'a' (the inner radius), reduced by the cortical thickness.

Torsional stress at fracture (T_m in N/m^2), where M is maximum torque capacity expressed in Nm, is given by

$$T_m = \frac{M \cdot b}{I_p}$$

Shear modulus (T_s in N/m^2), where k is stiffness expressed in Nm per radian twist and L is the length of the diaphysis between the castings, is calculated as:

$$T_s = \frac{k \cdot L}{I_p}$$

Statistics. The parameters obtained were plotted as a function of age of animal and the regression lines were calculated. For the most part, an estimated breaking point was selected for the calculation of two lines. Correlation analyses were used for some parameters, and prior to the experiment $p < 0.05$ was decided upon as significant.

Results

Macroscopic observations

All the bones tested were fractured by spiral type fractures, and in most cases there were one or two longitudinal components in the fracture path. When the torsion machine stopped immediately after the final fracture, some bones still had a thin cortical continuity somewhere along the fracture. This phenomenon was unrelated to age.

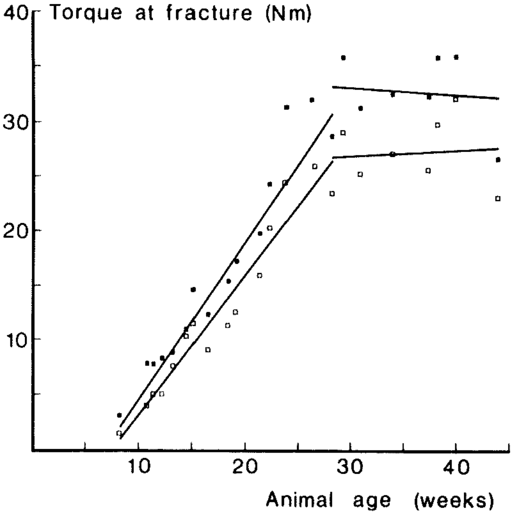

Figure 2. Ultimate torque (Nm) as a function of the age of the animal (weeks) for both tibiae (open squares) and femora (filled squares). For each bone type, two regression lines have been calculated, with the 28-week-and-2-day-old dog as a breaking value.

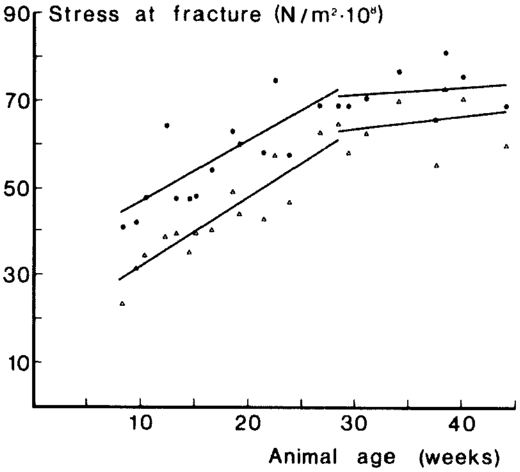

Figure 4. Stress at fracture ($N/m^2 \cdot 10^6$) for tibiae (open triangles) and femora (filled circles) as a function of animal age. For each bone type two regression lines have been plotted. Their breaking value is the 28-week-and-2-day-old dog.

Ultimate failure tests

Maximum torque capacity. Concomitant with the growth of the dogs, torque at final fracture increased for tibiae and femora to an age of approximately 30 weeks. The increase seems linear until that age, but the trend thereafter is not that apparent, as illustrated in Figure 2.

Stiffness. The torque-twist curve of all the bones tested had an initial linear part. The slope is an expression of the stiffness of the bone, and for both tibiae and femora it increases up to an age of approximately 30 weeks (Figure 3).

Growth and maturation. A commencement of epiphyseal closure was observed from an age of 14 weeks, but unfused epiphyses were still found at an age of 38 weeks. The correlation coefficients of the parameters expressing

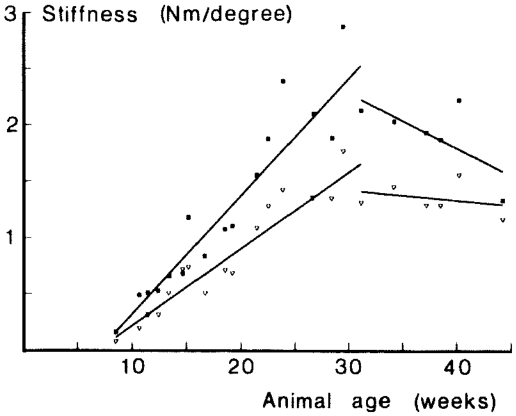

Figure 3. Stiffness for tibiae (open triangles) and femora (filled squares) as a function of animal age. For each bone type the value of the 31-week-old dog was chosen as the breaking point for two regression lines.

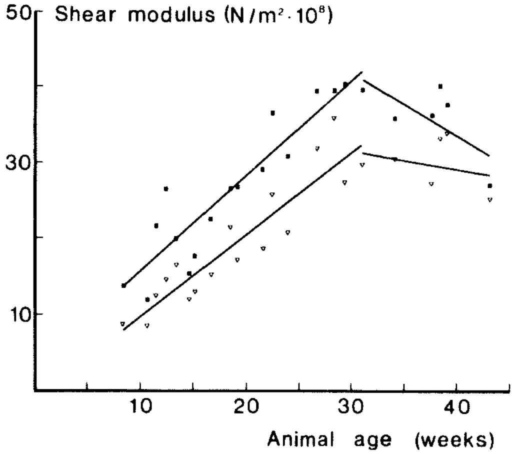

Figure 5. Shear modulus ($N/m^2 \cdot 10^8$) for tibiae (open triangles) and femora (filled squares) as a function of animal age. Two regression lines have been calculated for each bone type, and their breaking point is the 31-week-old dog.

growth, animal weight, bone length and bone weight, vary between 0.81 and 0.99. Animal weight ceased to increase at an age of around 30 weeks.

Stress and shear modulus. Stress at ultimate failure increased approximately three times for tibiae and twice for femora. The shear modulus for tibiae and femora increased especially during the first 25–30 weeks of life (Figures 4 and 5).

Deformation. Twist at final fracture did not change with age. The femora fractured at a twist angle of 16.75 to 28.0 degrees, and the tibiae at a twist angle of between 24.0 and 38.0 degrees. Twist at final fracture, expressed as a function of the length of the free part of the bone, was larger for the youngest bones, as illustrated in Figure 6. The relation between the length of the linear part of the deformation and the non-linear part did not change during growth of the dogs.

Load and unload tests

Stiffness. At all ages, stiffness of the bones decreased for every new load cycle. The quotient of stiffness for the first and the last load cycle was greatest for the bones from the young puppies (Figure 7).

Torque and twist. Torque at fracture was less for the load-unload-tested bone as compared with the paired bone which was uninterruptedly twisted to fracture, except for four pairs of tibiae and two pairs of femora. The difference is significant. Except for five pairs of tibiae and two pairs of femora, the ultimate twist was less for the load-unload-tested bone. The difference is statistically significant.

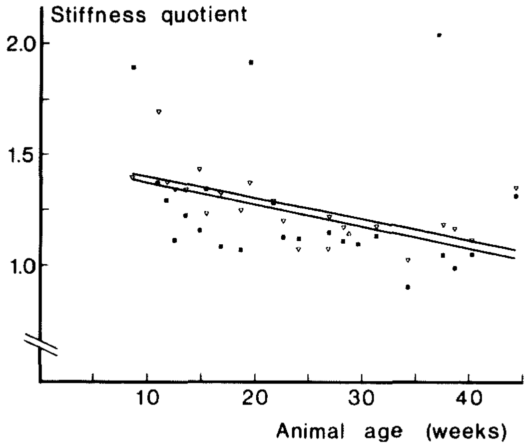

Figure 7. The stiffness quotient for the first and last load-cycle of the load-and-unload-tested bones plotted as a function of animal age. One regression line for tibiae (open triangles) and one for femora (filled squares) are plotted.

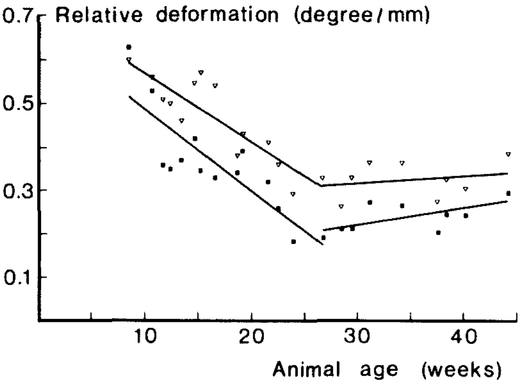

Figure 6. Utimate twist divided by twisted bone length (degree/mm) as a function of dog age. For both tibiae (open triangles) and femora (filled squares) two regression lines have been plotted with the breaking value from the 26-week-and-4-day-old dog.

Linear deformation. The length of the linear part of the torque-twist curve was shorter for the fracture load cycle than for the primary load cycle, in all the load-and-unload-tested bones.

Residual deformation. The residual deformation after load-unload tests varied between 0.90 and 4.39 degrees for femora, and between 0.87 and 6.27 degrees for tibiae. Apart from the two highest values, which both belonged to the youngest dog, there was no further age-variation. The residual deformation as a function of ultimate twist of the paired bone did not vary with age.

Discussion

Some mechanical studies on juvenile bone of varying age have been reported. Vinz (1970) made tension tests on machined specimens of femora, and found that the strength and stiffness of human bone increased from infancy up to 40 years of age. Currey & Butler (1975) studied femoral specimens from humans between 2 and 48 years of age in bending tests, and they also found an increase in strength and stiffness.

The increase of stiffness and torque at fracture of entire bones, as recorded in the present study, is due to an increased cross-section area (= increased polar moment of inertia), as well as to an increase in the strength parameters of the bone material, stress at fracture and shear modulus. These latter parameters increase during the first period of the dog's life concomitant with the growth of the young dog. This means that no important mechanical maturation as measured in bone strength was found after the cessation of growth in the dog.

Unlike Netz and co-workers (1979, 1980a and b), earlier investigators regarded the nonlinear deformation of diaphyseal bone as a consequence of its plastic structural properties (Burstein et al. 1972, Piekarski 1970). This should mean that unloading from within the non-linear part of the torque-twist curve would leave a permanent residual deformation. Previous load-and-unload tests of adult diaphyseal bones (Jonsson & Eriksson 1984) have shown a minor residual deformation which is mainly visco-elastic in its nature, i.e. it decreases with time. The load-unload-tested bones in the present investigation do not show any age-variation in the scale of residual deformation. A greater plasticity on the part of the juvenile bones, as described and discussed by Currey & Butler (1975) and Vinz (1970), thus cannot be verified. Furthermore, the decrease in stiffness of the load-unload-tested bones is of the same nature and of the same size at all ages. This supports the idea that juvenile bones, like adult bones, are mainly elastic-brittle and that non-linearity of the torque-twist curve is due to micro-cracking.

The results show that the deformation at ultimate fracture per twisted bone length unit is greatest for the youngest dogs. Contrary to the results of Currey & Butler (1975) and Vinz (1970), no relative decrease in the non-linear part of the deformation during growth of the bones was found. We consider that the discrepancy between the present results and those of Currey & Butler (1975) and Vinz (1970) is due to differing experimental models, chiefly to the important difference between fresh entire bones, as opposed to machined specimens, as test bodies.

The *linear* deformation relative to bone-size is greater for the younger bones, indicating a need for greater local deformation to start the micro-cracking. Also, the *non-linear* deformation compared to bone size is greater for the younger bones. These results show that micro-crack coalescence into final fracture requires a greater cortical deformation in younger bone compared to adult bone. A greater resistance to initiation and coalescence of micro-cracks in the juvenile bone, compared to the adult bone, expresses a greater toughness, which may be due to a different chemical or morphological composition (cf. Eriksson 1975).

Since the difference between the torque-twist curve parameters for entire paired bones is insignificant (Jonsson & Strömberg 1984), each dog has two identical test-pieces. In the present study, the one bone of a pair which was load-unload tested with a successively increasing twist angle for each load cycle, fractured at a lower torque and at a smaller twist angle than the paired bone, which was uninterruptedly twisted to fracture. This is also in accordance with visco-elastic properties, and may be due to a time-dependent stress-redistribution, which permits some cracking also during unloading of the bone (Jonsson & Eriksson 1984).

This study reveals several mechanical properties of diaphyseal bone which are without age-variation. However, some age-dependent mechanical properties have been noted. Mechanical alterations in growing bone have been ascribed by some authors to changed mineral content (Vose & Kubala 1959, Currey 1969), but by others also to changes in collagen content and collagen cross-linking (Vogel 1979, Ekeland et al. 1982). Elucidation of the supposed existence of these changes will be a future challenge.

Acknowledgements

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References

Blount, W. P. (1955) *Fractures in children,* p. 2. Williams & Wilkins, Baltimore.

Borden, S. (1975) Roentgen recognition of acute plastic bowing of the forearm in children. *Clin. Orthop.* **125**, 524–530.

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

Currey, J. D. (1969) The relationship between stiffness and the mineral content of bone. *J. Biomech.* **2**, 477–480.

Currey, J. D. & Butler, G. (1975) The mechanical properties of bone tissue in children. *J. Bone Joint Surg.* **57**-A, 810–814.

Ekeland, A., Engesaeter, L. B. & Langeland, N. (1982) Influence of age on mechanical properties of healing fractures and intact bones in rats. *Acta Orthop. Scand.* **53**, 527–534.

Eriksson, K. (1975) Fracture toughness and the distribution of inclusions. *Scand. J. Met.* **4**, 173–176.

Jonsson, U. & Strömberg, L. (1984) Uniformity in mechanics of long bones at torque. A dog experiment. *Acta Orthop. Scand.* **55**, 347–348.

Jonsson, U. & Eriksson, K. (1984) Microcracking in dog bone under load. A biomechanical study of bone visco-elasticity. *Acta Orthop. Scand.* **55**, 441–445.

Netz, P., Eriksson, K. & Strömberg, L. (1979) Nonlinear properties of diaphyseal bone. An experimental study on dogs. *Acta Orthop. Scand.* **50**, 139–143.

Netz, P., Eriksson, K. & Strömberg, L. (1980a) Material reaction of diaphyseal bone under torsion. An experimental study on dogs. *Acta Orthop. Scand.* **51**, 223–229.

Netz, P., Eriksson, K. & Strömberg, L. (1980b) Ultimate failure of diaphyseal bone. An experimental study on dogs. *Acta Orthop. Scand.* **51**, 583–588.

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

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

Vinz, H. (1970) Die Änderung der Festigkeitseigenschaften des kompakten Knochengewebes im Laufe der Alternsentwicklung. *Morph. Jb.* **115**, 257–272.

Vogel, N. G. (1979) Influence of maturation and aging on mechanical and biomechanical parameters of rat bone. *Gerontology* **25**, 16–23.

Vose, G. P. & Kubala, A. L. (1959) Bone strength – its relationship to X-ray determined ash content. *Hum. Biol.* **31**, 262–270.