Chapter A1
Cortical Bone

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A1.1 Composition

A1.1.1 Overall

The main constituents are the mineral hydroxyapatite, the fibrous protein collagen, and water. There is some non-collagenous organic material.

Highly mineralized bone (petrosal bones of some non-human mammals) has little organic material (8% in the horse petrosal to 3% in the tympanic bulla) [3]. (Almost certainly human ear bones will be somewhere near or in this region, though they seem not to have been studied.)

A1.1.2 Organic

The main organic component is collagen. Most is Type I, but there are small amounts of Type III and Type VI, found in restricted locations [4]. Slowly heated collagen shrinks at a particular temperature, giving an indication of the stability of the molecules. Bone collagen in men has a shrinkage temperature of about 61.5°–63.5°C up to the age of about 60, but about 60°C over that age. Bone from women showed much greater variability [5]. About 10% of the bone organic material is non-collagenous, mainly non-collagenous protein, NCP. The main ones are listed below. They have supposed functions that change rapidly.

- Osteocalcin (OC), or bone Gla protein (BGP)
- Osteonectin (ON), or SPARC
- Osteopontin (OPN) or secreted phosphoprotein I (SPPI)

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Table A1.1 Composition of Cortical Bone

	Water	Organic	Ash	Source
Mass %	12.0	28.1	59.9	[1]
Volume %	23.9	38.4	37.7	[1]
Volume %	15.5	41.8	39.9	[2]

Table A1.2 Density of Cortical Bone

Wet bone	1990 kg m^{-3} [1]

- Bone sialoprotein (BSP)

The relative amounts of these proteins can vary greatly. Ninomiya *et al.* [6] report far more osteocalcin (31 times) in cortical bone than in trabecular bone, and far more osteonectin (29 times) in trabecular bone than in cortical bone.

A1.1.3 Mineral

The mineral has a plate-like habit, the crystals being extremely small, about 4 nm by 50 nm by 50 nm. The mineral is a variant of hydroxyapatite, itself a variant of calcium phosphate: $Ca_{10}(PO_4)_6(OH)_2$ [7]. The crystals are impure. In particular there is about 4–6% of carbonate replacing the phosphate groups, making the mineral technically a carbonate apatite, *dahllite*, and various other substitutions take place [8].

A1.1.4 Cement line

The cement line round Haversian systems (secondary osteons) contains less calcium and phosphorus, and more sulphur than nearby parts of bone. This may indicate the presence of more sulphated mucosubstances, making the cement line viscous [9].

A1.2 Physical Properties

A1.2.1 Density

A1.2.2 Electromechanical behavior

Strained bone develops electrical potential differences. These used to be attributed to piezoelectric effects. However, the size of the piezoelectric effects is small compared with those produced by streaming potentials [10]. Furthermore, there were various anomalies with the potentials generated, which did not always accord with theory. The consensus now is that 'SGPs' (stress-generated potentials) are overwhelmingly caused by streaming potentials [10, 11]. Scott and Korostoff [12] determined, amongst other things, the relaxation time constants of the stress generated potentials, which varied greatly as a function of the conductivity and viscosity of the permeating fluid. As an example of their findings: a step-imposed loading moment which produced a peak strain of 4×10^{-4} induced an SGP of 1.8 mV, yielding a value of the SGP/strain ratio of 4500 mV. The SGP decayed rapidly at constant strain, reaching zero within about one second. For more detail, the complex original paper must be consulted.

A1.2.3 Other Physical Properties

Behari [10] gives a useful general review of many 'solid state' properties of bone, both human and non-human, many of which are not dealt with here. These properties include the Hall effect, photo-electric effects, electron paramagnetic resonance effects and so on.

A1.3 Mechanical Properties

A1.3.1 General

There is a great range for values in the literature for many reasons. Amongst these are:

(a) Different treatment of specimens Drying bone and then re-wetting it produces some small differences [13], as does formalin fixation [14]. Testing bone dry produces results quite different from those in wet bone; dry bone is stiffer, stronger, and considerably more brittle. Very small samples produce values for stiffness and strength less than those from larger samples [15, 16]. High strain rates generally produce a higher modulus of elasticity, a higher strength [17], and a greater strain to failure than specimens tested at low strain rate.

(b) Different age and health of donors Age may affect intrinsic properties. Osteoporotic bone may differ from 'normal' bone in ways other than the fact that it is more porous; there is evidence that the collagen is different from that in similar-aged non-osteoporotic subjects [18]. Bone from osteogenesis imperfecta patients has a higher proportion of Type III and Type V collagen compared with Type I collagen, than bone from normal subjects [19]. Bone collagen from osteopetrotic subjects is in general older than that from normal subjects, and has correspondingly different properties [5].

(c) Differences between bones, and sites in the bones The ear bones (ossicles) and portions of the temporal bones (petrosals) are highly mineralized, and will undoubtedly be stiffer and more brittle than others (though they seem not to have been investigated in humans). Long bones differ along their length and around their circumference. The distal femur is less highly mineralized and weaker in tensile and compressive static loading, and at any level the posterior part is similarly less mineralized and weaker [20].

The values reported below should be considered paradigmatic, that is, to be valid for a well-performed test on bone obtained from a middle aged person with no disease. Other values are reported in such a way as to make it clear how some property is a function of other features of the specimen.

A1.3.2 Stiffness

(a) General There are two ways of testing bone: mechanically by relating stresses to strains; ultrasonically, by subjecting the bone to ultrasound and measuring the velocity of the sound. From a knowledge of the density one can then obtain a stiffness matrix. If this is inverted it becomes a compliance matrix, the reciprocal of the individual terms of which are equivalent to the so-called technical moduli derived by mechanical testing [21]. Reilly and Burstein [22] give mechanical values, and Ashman *et al.* [23] give ultrasonic measurements. Reilly and Burstein [22] assumed transverse isotropy (that is, symmetry around the longitudinal axis of the bone), while Ashman *et al.* [23] assumed orthotropy (that is, that the values for stiffness could be different in the longitudinal, radial and tangential directions).

Reilly and Burstein [22] give values for Young's modulus at a number of intermediate angular orientations, but they do not form a very uniform set.

(b) Tensile modulus versus compressive modulus Reilly *et al.* [24] tested femoral specimens specifically to determine whether the value for Young's modulus was different in tension and compression. A paired Student's 't' test showed no significant difference between the compressive and tensile moduli at the 95% confidence level. Calculations on their data show the the 95% confidence interval ranged from compression modulus 1.72 GPa higher to tension modulus 0.27 GPa higher. The load-deformation traces showed no change of slope going from compression into tension and vice versa.

(c) Very small specimens The bending modulus of very small specimens was 6.62 GPa [5].

(d) Locational variations: Metaphysis versus diaphysis Young's modulus has been determined in three-point bending for extremely small plates (7 mm by 5 mm by (about) 0.3 mm) from the femoral metaphyseal shell and from the diaphysis of the same bones [16].

Table A1.3 Mechanical Properties

	Femur Tension [23]	Femur Tension [22]	Femur Compression [22]
Elastic moduli (GPa):			
E_1	12.0	12.8	11.7
E_2	13.4	12.8	11.7
E_3	20.0	17.7	18.2
Shear moduli* (GPa):			
G_{12}	4.5	–	–
G_{13}	5.6	3.3	–
G_{23}	6.2	3.3	–
Poisson's ratios:			
ν_{12}	0.38	0.53	0.63
ν_{13}	0.22	–	–
ν_{23}	0.24	–	–
ν_{21}	0.42	0.53	0.63
ν_{31}	0.37	0.41	0.38
ν_{32}	0.35	0.41	0.38

Subscript 1: radial direction relative to the long axis of the bone, 2: tangential direction, 3: longitudinal direction.

* Shear values are included under tension for convenience.

Table A1.4 Locational Variations in Modulus

Location	Longitudinal (GPa)	Transverse (GPa)	Source
Metaphysis	9.6	5.5	[16]
Diaphysis	12.5	6.0	[16]

The differences between these values and those reported by Reilly and Burstein [22] are probably attributable not to the difference in testing mode, since bending and tension tests from the same bone generally give similar values for Young's modulus, but to the very small size of the specimen, and to the rather low density of the specimens.

(e) Compression; effect of mineral The compressive behavior of cubes, relating the properties to the density of the specimens gives, using ρ_a (fat-free mass divided by anatomical volume, g cm^{-3}) as the explanatory variable:

Young's modulus (GPa) = $3.3\rho_a^{2.4}$ for compact bone [25].

The higher values of ρ_a were of the order of 1.8 g cm^{-3}(=1800 kg m^{-3}); this equation [25] predicts a value of 13.5 GPa for such a specimen. Multiple regression analysis showed that the dependence of Young's modulus on density was caused by

Table A1.5 Moduli of Osteons

Modulus (GPa)	Longitudinal Osteons	'Alternate' Osteons	Source
Tension	11.7	5.5	[26]
Compression	6.3	7.4	[27]
Bending	2.3	2.6	[28]
Torsional*	22.7	16.8	[29]

* Values for an 80-year-old man excluded.

the effect of porosity on density, and that, in these specimens, the effect of mineral content was insignificant.

(f) Single secondary osteons Ascenzi and co-workers [26–29] distinguish two types of secondary osteon: 'longitudinal' osteons, whose collagen fibres have a basically longitudinal orientation, and 'alternate' osteons, whose fibres have markedly different courses in neighboring lamellae. (This difference is a contentious issue.)

N.B.: These studies of Ascenzi and co-workers [26–29] are widely quoted, so beware of some apparent anomalies (apart from changes in nomenclature between papers). The bending modulus is remarkably low compared with the tension and compression moduli. The torsional (shear) modulus is remarkably high, compared both with the shear modulus values obtained by others (above), and with the tension and compression values. Torsional moduli are expected, on theoretical grounds, to be less than the tension and compression moduli. Furthermore, the large differences between the tension and compression moduli have not been reported elsewhere.]

(g) Strain rate effects Calculations [30], incorporating data from non-human as well as human material, predict that Young's modulus is very modestly dependent upon strain rate:

$$E = 21402(strain\ rate\ (s^{-1}))^{0.050}\ MPa$$

[N.B. statements about strain rate effects in bone are suspect unless it is clear that the workers have taken machine compliance into account!]

(h) Viscoelastic-damage properties Viscoelastic time constant (the value τ (s) in the equation):

$$\in(t) = \beta_1 \exp[t_o - t / \tau] + \beta_2$$

where the betas are parameters, t is time (s), t_o is time at which the specimen is held at a constant stress below the creep threshold: 6.1 s [31]. For reference, its value in bovine bone: 3.6 s.

Table A1.6 Strength of Cortical Bone [22]

Mode	Orientation	Breaking Strength (MPa)	Yield Stress (MPa)	Ultimate Strain
Tension	Longitudinal	133	114	0.031
	Tangential	52	–	0.007
Compression	Longitudinal	205	–	–
	Tangential	130	–	–
	Shear	67	–	–

Table A1.7 Locational Variations in Strength

Location	Longitudinal (MPa)	Transverse (MPa)	Source
Metaphysis	101	50	[16]
Diaphysis	129	47	[16]

Table A1.8 Strength of Osteons

Strength (MPa)	Longitudinal Osteons	‘Alternate’ Osteons	Source
Tension	120	102	[26]
Compression	110	134	[27]
Bending	390	348	[28]
Torsional*	202	167	[29]

* Values for an 80 year old man excluded.

A1.3.3 Strength

(a) Overall

(b) Combined loading Cezayirlioglu *et al.* [32] tested human bone under combined axial and torsional loading. The results are too complex to tabulate, but should be consulted by readers interested in complex loading phenomena.

(c) Metaphysis versus diaphysis Same specimens as reported for modulus above (Table A1.4) [16]. ‘Tensile’ strength calculated from the bending moment, using a ‘rupture factor’ to take account of the non-uniform distribution of strain in the specimen.

(d) Effect of mineral Keller [25], using the same specimens as above, provides the following relationship:

$$Strength = 43.9\rho_a^{2.0}\,(MPa)$$

[N.B.: The effect of mineralization, as opposed to density, is possibly of importance here; the original paper must be consulted.]

(e) Single secondary osteons The same nomenclature applies as for moduli of osteons (Table A1.5).

[N.B. The bending strengths and torsional strengths seem very high, even bearing in mind that no allowance has been made in bending for non-elastic effects.]

(f) Strain rate effects Bone will bear a higher stress if it is loaded at a higher strain (or stress) rate. Carter and Caler [17] found an empirical relationship that failure stress (σ_f (MPa)) was a function of either stress rate($(\dot{\sigma})$) or strain rate $(\dot{\in})$:

$$\sigma_f = 87(\dot{\sigma})^{0.053}$$

$$\sigma_f = 87\,(\dot{\in})^{0.055}$$

N.B. These relationships imply an increase of 44% in the failure stress if the stress rate is increased one thousandfold. This relationship has been found to be roughly the same in other, non-human, mammals.

(g) Creep Creep threshold (the stress below which no creep occurs): 73 MPa [31]. The equivalent value for bovine bone is 117 MPa [31]. Specimens in tension or compression were held at particular stresses [33]. The time (seconds) to failure is given as a function of normalized stress (stress/Young's modulus (MPa/MPa)):

$$\textit{Tension}: \quad \textit{Time to failure} = 1.45\times10^{-36}\ (\textit{normalized stress})^{-15.8}$$

$$\textit{Compression}: \textit{Time to failure} = 4.07\times10^{-37}\ (\textit{normalized stress})^{-17.8}$$

(h) Fatigue Some workers report the log of the number of cycles as a function of the applied stress levels, some report the log cycle number as a function of log stress levels, and some report log stress levels as a function of log cycle number. [The last seems wrong, since the applied stress can hardly be a function of the number of cycles the specimen is going to bear, but it is frequently used in fatigue studies. It is not possible simply to reverse the dependent and independent axes because the equations are derived from regressions with associated uncertainty.] The variation between the results for different testing modes is considerable.

Carter *et al.* [34] report on the effect of Young's modulus of elasticity and porosity in their specimens. They find that Young's modulus is positively associated with fatigue life, and porosity is negatively associated:

Table A1.9 Effect of Remodeling [35]

Property	Primary Osteons	Haversian Osteons
Tensile Strength (MPa)	162	133
Ultimate Strain	0.026	0.022
Young's modulus (GPa)	19.7	18.0

$$\begin{aligned}
\log\ N_f &= -2.05\ \log \Delta\sigma_o && (S.E.\ 0.599)\\
\log\ N_f &= -4.82\ \log \Delta\sigma_o + 0.186\ E && (S.E.\ 0.387)\\
\log\ N_f &= -2.63\ \log \Delta\sigma_o - 0.061\ P && (S.E.\ 0.513)\\
\log\ N_f &= -4.73\ \log \Delta\sigma_o + 0.160\ E - 0.029\ P && (S.E.\ 0.363)
\end{aligned}$$

where N_f: number of cycles to failure; $\Delta\sigma_0$: initial stress range (these experiments were carried out under strain control, so stress range decreased as damage spread and the specimens became more compliant); E: Young's modulus (GPa); P: porosity (%). Incorporating Young's modulus into the equation has a marked effect in reducing the standard error; porosity has a much less strong effect.

[N.B. Many workers normalize their data in an effort to reduce the effect that variations in Young's modulus have in increasing the scatter of the results.]

Choi and Goldstein [15] provide alternate, somewhat higher values.

(i) Effect of remodeling Vincentelli and Grigorov [35] examined the effect of Haversian remodelling on the tibia. The specimens they reported were almost entirely primary or Haversian, with few specimens having a scattering of secondary osteons. [Unfortunately they probably (it is not clear) allowed their specimens to dry out, so it is not sure that bone *in vivo* would show the same behavior. However, their results are similar to those found in nonhuman specimens.]

Additional Reading

Cowin, S.C. (ed.)(1989) *Bone Mechanics* Boca Raton: CRC Press.

A more rigorous, less chatty and less biologically, oriented approach than the following books by Currey and by Martin and Burr. The chapters on mechanics (2, 6 and 7), written by Cowin himself, are particularly authoritative.

Currey, J.D. (1984) *The Mechanical Adaptation of Bones* Princeton: University Press.

Out of print, new edition in preparation. Tries to deal with all aspects of mechanical properties of bone as a material and of whole bones. Not overly technical. Written from a general biological perspective, thus, does not concentrate on human material.

Martin, R.B. and Burr, D.B. (1989) *Structure, Function and Adaptation of Compact Bone* New York: Raven Press.

There are not many values of mechanical properties here, but the treatment of the biology of bone, and of fatigue of bone tissue, is excellent and the discussion of remodeling, although now somewhat out of date, is a very good introduction to this intellectually taxing topic.

Nigg, B.M. and Herzog, W. (eds)(1994) *Biomechanics of the Musculoskeletal System* John Wiley: Chichester.

Deals with many aspects of biomechanics, including locomotion, with an emphasis on human material. There is a full treatment of the measurement of many biomechanical properties.

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