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Excerpt
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Part I

The Osteochondral System

1 General Principles Relating to the Joint Tissues and Their Function

If there is one tissue in the body that deserves special recognition for its ability to perform extremely demanding load-bearing duties it is articular cartilage – that thin, glistening layer of compliant tissue covering the bone ends of our articulating joints. Articular cartilage functions so effectively in its role at the forefront of the joint system because of its unique physicochemical properties and its complex integration with the undergirding subchondral bone. And whereas the bone, at least in a straightforward mechanical sense, behaves very largely as a stiff elastic substrate, the overlying cartilage exhibits a much more varied set of mechanical properties and these can, in turn, render the junction between these two adjoining tissues especially vulnerable. This first chapter will provide a brief overview of some key mechanical principles relating to joint function followed by a detailed analysis of the structure of articular cartilage. Subsequent chapters will explore the structure and mechanical properties of the integrated cartilage–bone system.

1.1 The Regulation of Joint Stresses and Joint Friction

1.1.1 Contact Stress Reduction and the Importance of Compliance

Articular cartilage is a tissue we mostly take for granted until, by virtue of its partial or complete destruction, we enter that all-too-familiar world of debilitating joint pain. But why should the loss or breakdown of this highly compliant cartilaginous layer be the cause of such widespread suffering? Recourse to several quite straightforward mechanical principles can help answer this question.

First, the bone covered by the compliant articular cartilage layer is rigid by comparison and herein lies a major problem. Imagine a pair of typically profiled condyles making direct bone-to-bone contact under a state of simple compression (LH schematic in Figure 1.1a). Unless there is perfect contour matching across their entire surfaces the compressive load will be transmitted only across that smaller area of direct contact resulting in a concentration of stress that may well exceed the mechanical limits of the bone.

There is also a second level of mechanical risk, but on a much smaller scale. Less related to the joint contour it arises more from the natural microscopic undulations of the bone surfaces that would, without any intervening layers of cartilage, be in direct

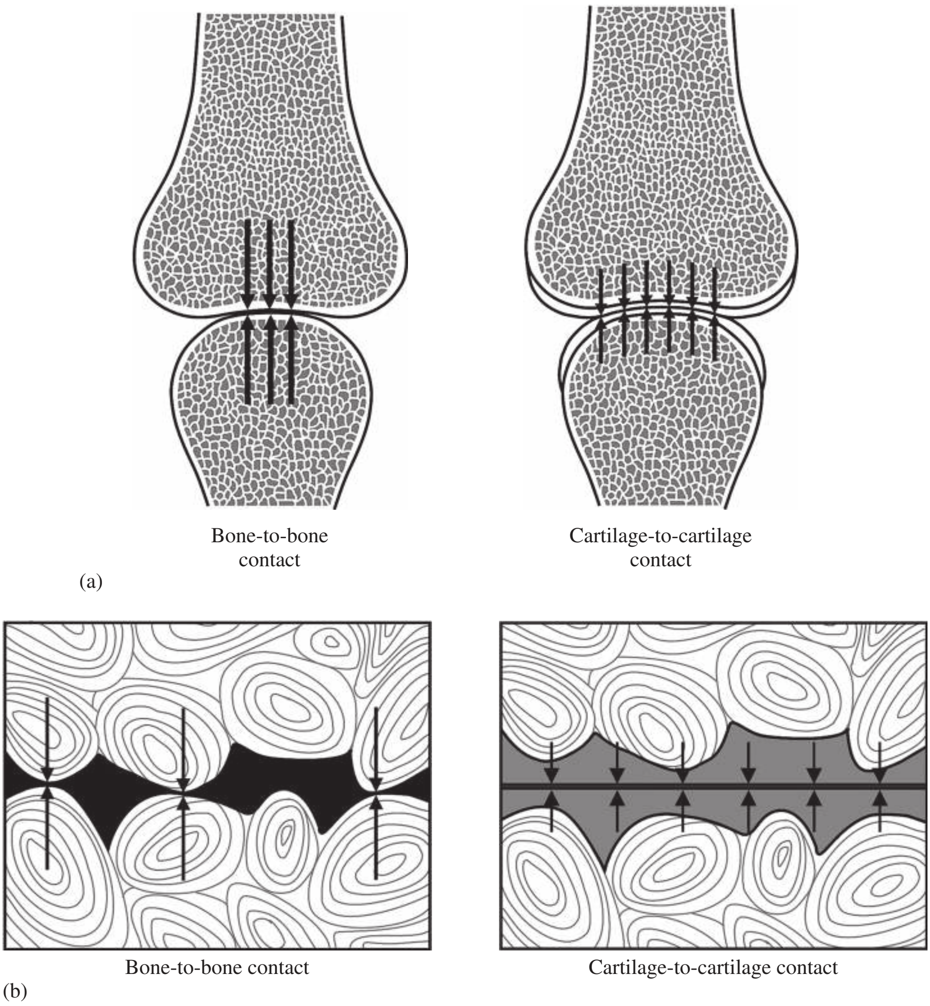

Figure 1.1 Schematics illustrating both macro-level (see a) and micro-level (see b) stress attenuation provided by the intervening layers of compliant articular cartilage (arrow length indicates approximate magnitude of stress). Images drawn by Samantha Rodrigues.

contact (LH schematic in Figure 1.1b). The load would again be concentrated at the high points of contact and sensed as pain in the innervated bone. The 'grinding' effect of repeated joint-loading and movement would lead, in time, to the destruction of the bone.

The elimination of these potentially damaging stress concentrations is, therefore, one of the primary biomechanical functions of the joint cartilage and is achieved by preventing the two rigid bone surfaces from having direct contact by means of the two compliant layers of articular cartilage (see RH schematics in Figures 1.1a and 1.1b). The latter deforms sufficiently to both maximise the area of contact between the differently contoured joint surfaces and eliminate these localised, more micro-scale contact points. The applied compressive load is now distributed over a much larger area thereby reducing the contact stresses transmitted into the bone tissues to a safe level.

Broom and Oloyede (1993) were the first to experimentally investigate this contact stress-attenuating role of articular cartilage in a study in which they glued layers of articular cartilage to a photo-elastic epoxy resin substrate, the latter acting as a model analogue of the rigid subchondral bone. By analysing the photo-elastic fringe patterns obtained under both quasi-static and dynamic compressive loading they were able to determine the levels of shear stress generated subchondrally with respect to depth below the compliant–rigid junction and compare the effectiveness of the articular cartilage in reducing these shear stresses under conditions of both quasi-static and dynamic compression. They inferred from these experiments that while articular cartilage provides a significant level of subchondral bone protection under both quasi-static and dynamic loading, greater bone protection is provided closer to the cartilage–bone junction under quasi-static than under dynamic loading conditions. They interpreted this difference in terms of the contrasting deformation mechanisms operating in the articular cartilage matrix at low versus high rates of loading. We explore this rate effect in more depth in Chapter 3 (see Section 3.2).

We can see the damaging consequences of high-stress concentrations (and their prevention) when two flat sheets of glass are brought together but between them a small fragment of grit has been inadvertently trapped. The consequences are all too obvious: the destructively high contact stresses at the grit site will quickly scratch or damage the glass. But simply by sandwiching several soft layers of paper (analogous to cartilage) between the two hard sheets the problem is averted. In fact, glaziers always cut their sheets of glass on a flat, soft cloth-covered surface for this very reason.

1.1.2 Lubrication Mechanisms in the Articulating Joint

The stiff subchondral bone is therefore protected from high concentrations of stress by its compliant covering of articular cartilage, but what then protects the 'protector'? The constant articulations performed by a joint over a lifetime would surely put at risk even its cartilage were it not for the very special conditions that prevail at the joint surfaces. A key function of the joint cartilage is therefore the provision of near friction-free movement and it can perform this critical role by virtue of it having a coefficient of friction of around 0.01 (Charnley 1960). By way of comparison polytetrafluorethylene ('Teflon') bearing materials have a coefficient of the order of 0.04 (McCutchen 1962a, 1962b).

McCutchen (1990) has provided an insightful overview of the history of synovial joint lubrication as well as highlighting the conflicting ideas and theories that have shaped our understanding of this important area. As a way of doing justice to the topic of joint lubrication within the confines of this short section we shall draw substantially on material presented in McCutchen's informative overview.

The science of joint lubrication began to develop with the very early suggestion of MacConaill (1932) that wedge-shaped synovial fluid-filled spaces were created between the articulating surfaces that were not completely congruent throughout their entire range of movement. MacConaill argued that pressure is then generated in this wedge of fluid as a result of the relative movement between the two surfaces. It is this pressurised-fluid film that keeps the two surfaces separated, supports the loading applied to the joint and facilitates very low friction articulation: in effect he was describing the well-known mechanism of hydrodynamic lubrication. It should be noted that the hydrodynamic mechanism proposed by MacConaill required relative movement between the two joint surfaces. However, in human-made bearing systems the same end goal of physically separating two surfaces by means of a pressurised film of fluid can also be achieved without any relative motion – simply by employing an external pressure pump to inject a lubricant directly into the bearing as is the case in the modern motorcar engine – a mechanism obviously not applicable to the living joint!

Following MacConaill (1932), others were able to show that the viscosity of synovial fluid decreased with an increasing rate of shear, a property change again consistent with a mechanism of hydrodynamic lubrication operating in the joint such that with higher speed motion a lower viscosity would still ensure effective lubrication without increasing the drag forces (Ropes et al. 1947; Ogston and Stanier 1953). Charnley (1960), however, challenged the hydrodynamic theory on three counts arguing that: (i) in some joint systems there is extended intimate contact between the two loaded surfaces such that the wedge space profile would not exist; (ii) where there is 'reciprocating' movement the pressurised fluid wedge, once generated by movement in one direction, would then be destroyed with any reverse movement; and (iii) hydrodynamic lubrication is not easily achieved with slow-moving surfaces under heavy loads. In fact, the synovial joint exhibits low friction with zero sliding between its surfaces (McCutchen 1962b). Charnley (1960) promoted, instead, the idea of the articular surface being inherently slippery, with low-friction joint articulation being primarily dependent on a mechanism of boundary lubrication. However, he did not rule out the possibility of there being a component of quasi-static hydrodynamic lubrication arising from the free synovial fluid present in the joint cavity.

In response to the idea of boundary lubrication McCutchen (1959, 1962a, 1962b) suggested that it might have been premature of Charnley to dismiss the role of fluid pressure in favour of some kind of inherent 'Teflon-like' boundary layer slipperiness created by an interaction between the cartilage surface and synovial fluid. Building on the idea of a 'hydrostatic bearing' McCutchen developed the concept of 'weeping lubrication' based on the fact that articular cartilage is both porous and deformable. He conducted a series of ingenious friction experiments using a closed-pore sponge material in which one face had been cut to expose its cells and was able to demonstrate that its slipperiness when wetted and loaded against a glass surface was a direct consequence of the pressurised fluid being trapped in the tiny pockets now sealed against the impervious surface. The bulk of the applied load was carried in a largely frictionless manner and as long as the fluid remained trapped McCutchen was able to show that the low-friction state persisted – gradual leakage over time brought the solid material of the sponge increasingly into contact with the glass surface, thereby increasing the friction.

McCutchen (1962a, 1962b, 1990) carried out similar experiments on articular cartilage, treating it as a 'fine-grained sponge' material possessing an ultra-low permeability, and demonstrated again that it is slippery when hydrated, but, as the fluid is squeezed out of its ultrafine pores, the friction increases. He argued that loading would pressurise the fluid in the pores almost to the level of the applied load and therefore carry most of this load with little left to be carried by direct solid-to-solid contact between the two cartilage matrices. McCutchen referred to this mechanism as 'weeping lubrication' in order to emphasise that fluid initially present between the two surfaces would not be responsible for the persisting low-friction contact between them. Rather, the fluid carrying the load is derived from within the hydrated cartilage matrix itself. He also reasoned that because the solid components of the two matrices would only be in very light contact – a consequence of the small component of the total load being carried by them – their sliding over each other will be lubricated successfully by the synovial fluid, in effect providing an additional minor component of boundary lubrication (McCutchen 1983).

That there is stress-sharing between the interstitial fluid and the solid components in the compressed articular cartilage matrix, and that this sharing changes over time, was first demonstrated experimentally by Oloyede and Broom (1991, 1993, 1994a, 1994b, 1996) and subsequently by Soltz and Ateshian (1998, 2000) and Park et al. (2003). Oloyede and Broom were able to show that a level of pore pressure was developed in the compressed articular cartilage matrix that approached the applied stress and was, thus, entirely consistent with McCutchen's mechanism of weeping lubrication. They were also able to demonstrate experimentally the transient nature of this hydrostatic pore pressure development, showing that it attained a maximum initial value (the maximum excess pore pressure) and with sustained loading it decayed gradually to a near-zero level over a period of several hours. This indicated that the applied load was being progressively transferred from the fluid phase into the solid components of the matrix via a classical consolidation mechanism.

As has been pointed out in the review by Ateshian (2009), now that the pressurised interstitial fluid has been shown to facilitate one of the primary friction-reducing mechanisms in the joint, by allowing this interstitial pressure to subside (by matrix consolidation) it becomes possible to independently investigate the effectiveness of other mechanisms and especially that provided by boundary lubricants. With sufficient relative motion between the articulating surfaces there is also hydrodynamically induced fluid film lubrication to add to the mix of mechanisms now recognised as contributing to the joint's low-friction function. The science of joint lubrication is a still-developing field; the literature is vast and growing, and interested readers are referred to reviews by Neu et al. (2008), McNary et al. (2012) and Daniel (2014).

1.2 The Structural Meaning of Elastic Stiffness

The mechanical property termed elasticity refers to a material's ability to recover completely from any deformation of its structure and it will be obvious that elasticity has its limits – if we stretch or compress too much we exceed the material's elastic limit and either irreversible deformation or fracture will be the end result.

Conventional materials such as most minerals, metals, ceramics and glass, etc., all derive their relatively high stiffness from the balance between the attractive and repulsive forces that bind their component atoms or molecules into what is usually (but not always) an ordered or crystalline arrangement. In effect, the distance separating the atoms or molecules defines an interaction of lowest potential energy, or greatest stability. The distances over which the fundamental forces act are very small so the degree to which these materials can be elastically deformed is very small and directly related to the permissible amount of stretching or compressing of the interatomic or intermolecular bonds before they are either severed or the component atoms or molecules are irreversibly translated into new equivalent lowest energy positions as occurs in permanent or plastic deformation. Importantly, any elastic stretching or compressing results in an increase in strain energy of the material and it is the return to the lowest energy state on removal of the deforming stress that is the driving force for its near-instantaneous elastic recovery.

For a conventional metal the limit of elastic stretching or compressing before irreversible changes occur is typically less than ~0.2 per cent of its undeformed dimensions. Most metals possessing a degree of ductility will be permanently or plastically deformed if strained beyond this value, whereas highly brittle solids such as many minerals, ceramics and inorganic glasses may reach a slightly higher strain and then simply fracture. Whether this elastic behaviour ends in either permanent deformation or brittle fracture, it comes under the general definition of conventional, low-strain elasticity.

1.3 Fundamental Principles Governing Compliant Versus Stiff Tissues

The osteochondral junction is responsible for integrating structurally the articular cartilage and subchondral bone with their contrasting mechanical properties to create a highly successful load-bearing system. The subchondral bone is required to function mechanically as a relatively rigid elastic material and can, therefore, be categorised *approximately* as having small-strain elastic properties.

The binding forces responsible for this high elastic stiffness reside mostly in the bone's brittle mineralised component, namely the calcium hydroxyapatite in which the amount of elastic stretch is defined by the strength of intermolecular bonds within its complex crystalline structure. The less stiff collagen fibrils embedded and constrained within it act primarily as a highly structured tensile-reinforcing component that reduces the risk of fracture occurring in the mineralised phase under tensile loading conditions and thereby increases the bone's toughness. We shall explore in more detail this important concept of toughness as applied to fracture of the osteochondral junction in Chapter 4.

Although we have described the collagen fibre or fibril as the less stiff component in the subchondral bone, such a statement requires some careful qualification. First, the collagen fibril, by virtue of its inherent structure, would be expected to fit more within the category of small-strain elasticity, although at first glance this might appear to contradict what we observe experimentally. The assembly of the collagen fibril begins with a primary α-helical polymer chain developed from a repeating sequence of amino acids. Then, through several sub-levels we arrive finally at the quarter-staggered, over-lapping array of tropocollagen molecules to form the collagen fibril (see reviews of collagen structure by Nimni and Harkness 1988; van der Rest and Garrone 1991), a hierarchical structure that is highly ordered and effectively 'bond-locked', with limited ability to extend axially before irreversible structural damage results.

Rigby et al. (1959) suggested that for the rat tail tendon, strains beyond about 4 per cent result in the rupture of secondary bonds, but not necessarily the primary bonds, of the collagen molecules. Harkness (1961) in his extensive review of collagen described the fibril as being relatively inextensible with actual rupture in the collagen *fibre* occurring beyond strains of 10–20 per cent. Kastelic and Baer (1980) report irreversible elongation (they termed this 'yield') of tendon fibres with strains from 2 per cent and upward depending on maturity. Earlier experiments by Hall (1952) demonstrated that at a normal pH of 7.0 collagen fibre elasticity is due primarily to internal energy changes arising from the stretching of bonds and bond angles rather than to entropy changes, the latter being the case for both elastomeric rubbers and the elastin fibre, a topic we shall discuss in some depth in Chapter 7.

The elastic strain limit of the collagen fibre is still considerably higher than that of many conventional crystalline solids (typically ~0.2 per cent strain) but this reflects the much more complex multi-level nature of the bonding in collagen's hierarchical structure (involving hydrogen bonds, electrostatic interactions and covalent bonds) in contrast to the regular interatomic bonds that determine the elastic limit of a typical crystalline metal. It is important to emphasise, too, that while collagenous arrays such as those comprising tendons, ligaments, and the disc annulus are able to exhibit elastic strains as high as 10–15 per cent (see e.g. Rigby et al. 1959; Baer et al. 1988) this is not a reflection of their inherent material elasticity. Rather, such large elastic strains are derived from their crimped morphology and its straightening under a tensile load. Once this 'geometric' crimp is eliminated any further elastic stretch is limited to that able to be derived from the fibril's bond-locked structure.

For the collagen fibril or fibre we, therefore, need to distinguish between an intrinsic bond-based elasticity (inherently low strain and probably less than ~4 per cent) and a geometric or configuration-based elasticity which has the ability to generate very large elastic strains that still harness the intrinsic bond-based interactions. A nice example illustrating this principle is a coiled steel spring: the intrinsic elastic stiffness of steel is ~200 GPa and yet we can have a spring constructed from this same steel having vastly lower stiffness values by virtue of the elastic uncoiling of its coiled configuration. The material stiffness of the steel remains unchanged throughout the elastic unravelling of the coils.

Although the articular cartilage matrix is relatively rich in collagen, for it to function effectively in its stress-attenuating role it is required to be highly deformable in a completely recoverable sense and yet still act in the 'front line' of joint loading.

Cartilage derives these important mechanical properties from a highly complex set of structural and physicochemical relationships between its constituent components. So, what is the structure of articular cartilage and how does this structure give rise to the tissue's functional properties?

1.4 Composition of Articular Cartilage and Its Physico-Chemical Implications

In purely constituent terms, about two-thirds of the dry weight of mature articular cartilage is collagen and this is mostly type II collagen fibrils (Eyre 2002). The other major biochemical components are the proteoglycans – a broad class of macromolecules consisting of a protein core to which are attached glycosaminoglycans – the dominant one being aggrecan (Heinegard 2009). Typically, more than a 100 negatively charged chondroitin sulphate molecules bind to the aggrecan chains (Knudson and Knudson 2001). Two related physico-chemical properties arise from the macromolecular configuration characterising the proteoglycans. First, due to the repulsive forces between the fixed negative charges along the glycosaminoglycan chains they adopt an extended, rather than contracted, configuration and, thus, present as a potentially large open macromolecular structure. Second, the fixed negative charges will tend to be neutralised by counter ions which increase the chemical species concentration and results in water being drawn osmotically into the cartilage matrix.

The above two effects therefore give the cartilage matrix a very large water-binding potential such that more than 60 per cent of its wet weight can consist of matrix fluid. The swelling pressure generated by this in-drawing of fluid is countered by the unique fibrillar architecture of the healthy articular cartilage matrix and confers on it a range of biophysical and biomechanical attributes fundamental to its primary load-bearing function. The reader is referred to pioneering studies by Maroudas and co-workers in which the fundamental physico-chemical principles are developed that correlate composition with both the swelling and permeability properties of the cartilage matrix, both normal and degenerate (Maroudas 1968; Maroudas and Bullough 1968; Maroudas et al. 1968; Maroudas 1976).

1.5 Early Structural Models of Articular Cartilage

In a larger mammalian joint the articular cartilage covering the bone ends is typically between 1 and 2.5 mm in thickness with its depth conventionally divided into a number of relatively distinct structural zones before the bone substrate is reached. The earliest investigation of the relationship between the fine structure of articular cartilage and its functional role was conducted by the Scottish physician and anatomist William Hunter (Figure 1.2). In his paper *Of the Structure and Diseases of Articulating Cartilages* (Hunter 1742) which he read to a gathering of The Royal Society of London in 1743, Hunter's opening paragraph should bring delight to the eyes of any

Of the Structure and Diſeaſes of Articulating Cartilages, *by* William Hunter, *Surgeon.*

Read June 2. 1743.

THE Fabric of the Joints in the Human Body is a Subject ſo much the more entertaining, as it muſt ſtrike every one that conſiders it attentively with an Idea of fine mechanical Compoſition. Where-ever the Motion of one Bone upon another is requiſite, there we find an excellent Apparatus for rendering that Motion ſafe and free: We ſee, for Inſtance, the Extremity of one Bone moulded into an orbicular Cavity, to receive the Head of another, in order to afford it an extenſive Play. Both are covered with a ſmooth elaſtic Cruſt, to prevent mutual Abraſion; connected with ſtrong Ligaments, to prevent Diſlocation; and incloſed in a Bag that contains a proper Fluid depoſited there, for lubricating the Two contiguous Surfaces. So much in general.

Figure 1.2 Portrait of eighteenth-century Scottish physician and anatomist William Hunter (image from Wikipedia) together with a quote from the paper that he read to a meeting of the Royal Society of London on 2 June 1743.

modern joint biomechanist (Figure 1.2). It alludes most elegantly to the mechanical principles describing how the cartilage protects the bone-ends and to the requirements of both joint lubrication and stability.

As we will see shortly, Hunter was the first to provide us with a glimpse of the basic architecture of articular cartilage. He describes the fibres in articular cartilage as rising up perpendicularly from the underlying bone, likening its texture to the pile of velvet attached to its base of woven cloth. Although unable to view them with his ‘glass’, Hunter argued for the presence of transverse fibres to connect the perpendicular elements and thus form a ‘whole solid body’. Further, he refers to a fine covering membrane ‘firmly braced upon the surface’, this being exceedingly fine but readily demonstrated when the cartilage is macerated.

As noted by Clarke (1971), the Swedish anatomist Vilhelm Hultkrantz in 1898 developed the idea that the fibres in the surface layer of articular cartilage possessed strongly directional properties. He made multiple punctures across the articular surfaces of a variety of joints with a small round awl and demonstrated sweeping patterns of directional splitting which he interpreted as tracking the fibrous alignment in the surface layer. The image in Figure 1.3 illustrates this directional splitting tendency in the articular surface of a cartilage–bone sample that had been incrementally loaded in compression using a transparent indenter until rupture occurred. The sample was rehydrated and then pin-pricked and inked to reveal how the rupture had tracked along the contour of the pin-prick splits.