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Design and Biomechanics of the Oxford Knee

The description of the Oxford Knee starts with an explanation of the function of mobile bearings in knee prostheses. An obvious advantage is that the areas of contact between the joint surfaces are maximised. In this chapter, we shall show that wear at the polyethylene surfaces is thereby minimised and that optimal kinematics can be achieved with minimal risk of loosening. We will discuss the biomechanics of the cementless components and problems that may occur with the tibia.

Designing against wear

Articular surface shapes and contact pressures

Most surface replacements of the knee, total as well as unicompartmental, have articular surfaces like those shown in Figure 2.1, approximating to the shapes of the ends of the human femur and tibia. The metal femoral surfaces are convex and the polyethylene tibial surfaces are flat or shallowly concave. These shapes do not fit one another, in any relative position, and so only parts of their articular surfaces are in contact and able to transmit load.

Most prosthetic femoral condyles attempt to mimic nature and are polyradial, with the shortest radius posterior. Thus the area of contact is smaller in flexion than in extension (Fig. 2.1). However, the compressive loads transmitted across the interface are potentially greatest in flexion, attaining up to six times body weight during stair ascent and descent [1]. For a given load, the average contact pressure (load per unit area) at the articular surfaces is inversely proportional to the area of contact; therefore the less congruous the surfaces, the higher is the average pressure at their interface. The wear rate of ultra-high-molecular-weight polyethylene (referred to hereafter as 'polyethylene') is said to increase exponentially with increasing contact pressure, rather than linearly as would be expected from classical wear theory [2]; conversely, wear rate has been found to decrease with increasing contact area [3].
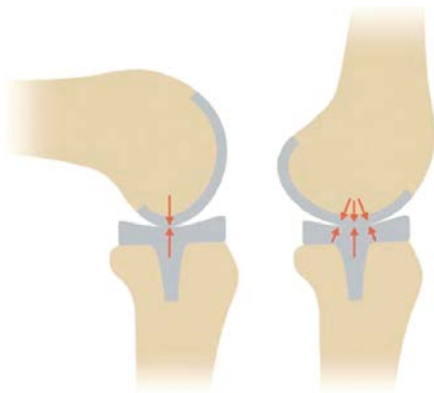
Figure 2.1 Typical polycentric incongruous knee replacement with a smaller contact area in flexion.

The natural knee

The presence of the cartilaginous menisci in the knee of humans (and of all other mammals) gives rise to an entirely different regime of contact (Fig. 2.2). Instead of one incongruous interface, two congruous interfaces are created, with much better distribution of load.

Fairbank, in 1948, first deduced that the human meniscus had a load-bearing function and suggested the mechanism of load transmission shown in Figure 2.3 [4]. The menisci consist mainly of collagen fibres disposed circumferentially to withstand the tensile hoop stresses engendered by load bearing; these stresses are resisted at the anterior and posterior horns by their attachments to the tibia [5]. The proportion of load transmitted indirectly by the menisci in human (and animal) joints has been estimated as between 45 and 70% of the applied load [6]. The remaining 30 – 55% is carried by the articular cartilage of the femoral and tibial surfaces within the embrace of the meniscus through their direct contact in the middle third of each plateau.
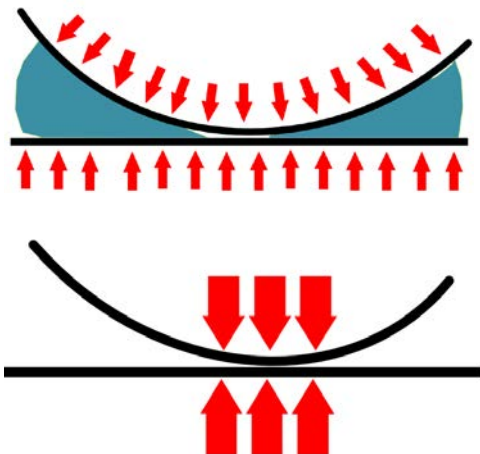

Figure 2.2 Load-sharing function of the meniscus, increasing effective contact area and reducing contact pressure. Loss of a meniscus reduces contact area and increases contact pressure.
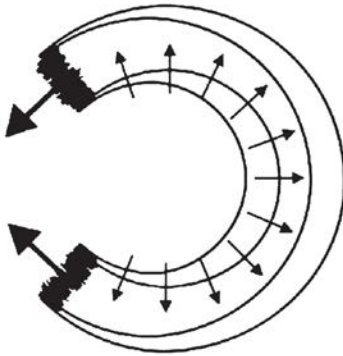
Figure 2.3 Mechanism of load transmission: the radially outward component of applied pressure is resisted by hoop stresses in the circumferential fibres of the meniscus. (Adapted from and reproduced with permission from Lippincott Williams & Wilkins [Shrive NG, O'Connor JJ and Goodfellow JW. Load-bearing in the knee joint. *Clin Orthop* 1978; **131**: 279–87].)

Mobility of the natural meniscus

Anteroposterior movements of the femoral condyles on the tibia during flexion–extension and axial rotation (Fig. 2.4) have to be accommodated by movements of the menisci. In 1680, Borelli [7] noticed that '*they are pulled forward when the knee is extended and backwards in flexion*'. Various estimates and measurements of these movements during flexion have been reported: 6 mm medially and 12 mm laterally [8]; 5.1 mm (SD 0.96) medially and 11.2 mm (SD 2.29) laterally [9]; medial anterior horn 7.1 mm (SD 2.49), medial posterior horn 3.9 mm (SD 1.75), lateral anterior horn 9.5 mm (SD 3.96), and lateral posterior horn 5.6 mm (SD 2.76) [10]. Freeman and his group suggest that the knee is a medial pivot with no movement medially; however, even Freeman's data suggests that there is movement of about 8 mm (Fig 2.4).
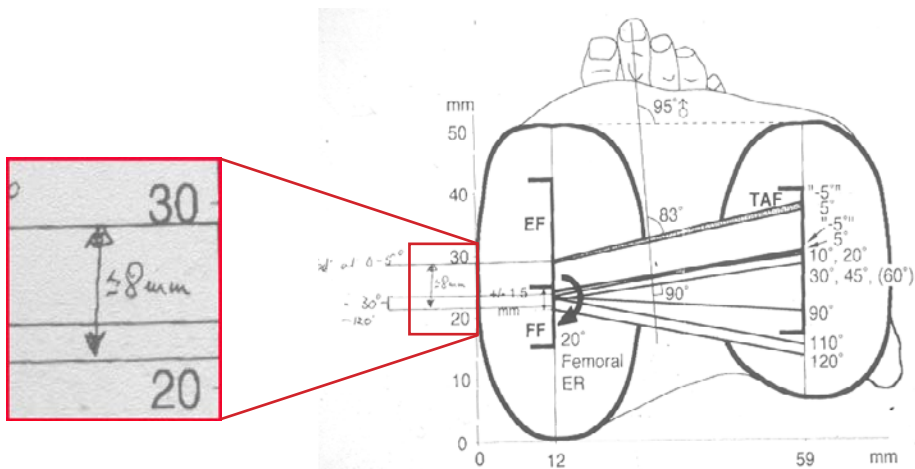

Figure 2.4 Anteroposterior movements of the femoral condyles on the tibia during flexion–extension and axial rotation, with annotations by Mr Michael Freeman.